非対称歩行環境での歩行適応に貢献する肢体運動の解明

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【目次】

1．要旨

本研究は脳卒中片麻痺者の歩行適応動態を分析することで、非対称歩行環境への歩行適応における肢体運動の影響を明確にすることを目的に行なった。目的達成のため、踵接地期、動的な力学的安定性、関節可動域の影響の 3 点を考慮し、split-belt treadmill 歩行課題を行った。まず歩行周期のうち踵接地期における全身体と接地下肢との力学的安定性を分析した。その結果、CoM-CoP 角（Centre of Mass-Centre of Pressure）がベルト速度の非対称な状態でも左右で対称化する適応的変化を示し、全身体と前脚の動的に安定性の高い相対的位置関係であることを明らかにした。次に、肢体運動における物理的な運動制限の影響を鑑別した。片側の一関節の一方向に作った軽微な運動制限であっても肢体間協調性に悪影響を及ぼす結果となった。しかし、CoM-CoP 角の適応的変化には影響がなかった。最後に、片麻痺において CoM-CoP 角の適応的変化を示すか否かでサブグループ化した結果、時空間変数の対称化における患者特異性を区別することができた。つまり、split-belt treadmill 歩行課題がもたらすベルト速度の左右非対称な歩行環境に歩行を適応させる上では、接地において力学的安定性の高い全身との相対位置に前脚を予測的に配置することができることが片麻痺者の歩行適応能力を決定する因子であった。以上のことから、非対称歩行環境での歩行適応に対して肢体間の協調運動の目的は、ステップ長の対称化が本質ではなく、動的安定性を通常の対称歩行と同様に収束させることであると考えられる。これは、split-belt treadmill を歩行運動学習の介入ツールとして応用する目的と、適用対象の解明に貢献する可能性がある。

2. 発表論文

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2. Verification of the adaptive parameters of the relative positions of the leading leg and the whole body at foot contact during split-belt treadmill walking

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   Proceedings of IEEE/SICE International Symposium on System Integration, 2019

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3. Adaptive changes in foot placement for split-belt treadmill walking in individuals with stroke

   K. Hirata, H. Hanawa, T. Miyazawa, K. Kubota, M. Sonoo, T. Kokubun, N. Kanemura

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3. 研究背景

3.1. 歩行障害の社会的背景

2016 年に発表された高齢社会白書[1]によれば、我が国では 65 歳以上の高齢者は過去最高の 3392 万人となり、総人口に占める割合も 26.7%と過去最高となっている。日常生活に介護を要する要介護（要支援）認定者数は 2013 年には 569.1 万人にのぼる。そのうち、全ての介護度において、認定の要因となった最たる疾患は脳卒中である。脳卒中は罹患者が 100 万人を超え、生活習慣病と関連した国民病と言える。さらに要介護状態になる要因第一位となるほど重篤な後遺症「片麻痺」を引き起こすために、片麻痺者の歩行自立度の維持、向上と介助者の負担軽減は、超高齢社会に突入した我が国において社会的解決課題に他ならない。学術領域に課せられた課題は、神経学的理論背景に基づく介入「ニューロリハビリテーション」の確立である[2]。そのためには、臨床において対象者に特定の介入方法を実施し、科学的手続きによって、中枢神経系を再組織化した神経学的機序の観点からその介入の有効性について説明することが必要である。また、それと同時に、ヒトの歩行の神経制御機構を明らかにするための基礎的な研究が求められる。こうした研究の成果は、介入方法の理論的背景を構築したり、改善効果を解釈するための基盤的な知見となる。

3.2.歩行の神経制御

ヒトの二足直立歩行、および四足動物の移動運動はロコモーション（Locomotion）と総称される。ロコモーションの中枢神経系による制御は、支配領域と歩行に果たす役割として 3 つに大別される[3]。1 つは大脳皮質、大脳辺縁系など高次脳機能が司る発動系で、歩き出しに関与す

るとされる。大脳皮質からは随意的な信号により駆動される随意的プロセスが発現され、その信号は皮質線条体投射や皮質脳幹投射、そして外側皮質脊髄路系を介して皮質下構造に伝達される。辺縁系や視床下部から脳幹への投射系が担う情動行動は、主に行動を誘発する信号の種類にかかわらず歩行動作や筋緊張の亢進など定型的な運動パターンが誘発される特徴を持つ[4)]。

第2に、小脳、大脳基底核が歩行の適応に関与している調節系である。四肢の筋緊張の制御と肢体運動の位相制御を運動感覚のフィードバック情報（入力）と大脳皮質からの運動指令（出力）を比較することで同時並列に歩行運動を調節すると考えられている[5)]。大脳基底核は大脳皮質からの指令を受け、再び大脳皮質へ戻すループなど異なる複数の経路を有している。基底核は歩行の調節、および発達を含めた学習に関与しているとされている[6)]。

3つ目は中脳を含む脳幹、脊髄が司る実行系である。上位脳による随意的な発動後に、これら下位脳が四肢の屈伸運動や左右肢の交互運動が組み合わさったサイクリックな全身運動がリズミカルに自動化する[7)]。脳幹中脳以下を残して、大脳との連絡を断った除脳ネコがトレッドミル上でのロコモーションを維持[8)]、また速度変化に歩調を合わせることが可能である[9)]ことをもって、脊髄内にリズム発生器（CPG、Central Pattern Generator）[10)]が内在し、歩行の維持に関わっていることが説明され、概念化されてきた。

つまり、歩行が随意的に開始されたのち、半自動的に維持し（実行系）、歩行環境の変化に合わせて適応すること（調節系）は大脳皮質を除く中枢神経系の下位レベルで可能である。

3.3. Split-belt treadmill

Split-belt treadmill は新規の歩行環境に対する学習（調節系）とパターン形成（実行系）の過程を明らかにする手法として用いられる。Split-belt 課題とは、歩行中に左右で分離したベルトの一側を予告なく変速させるもので、歩行を維持しながら、また円滑な肢体運動に適応する過程をもって歩行の運動学習を記述する手法である。これまで Dietz ら、Bastian らのグループを中心に、様々な条件と対象者に対して活発に研究が行われてきた [11-18]。下肢を中心とした肢体運動は協調的に変調することで歩行環境に適応していく。適応の過程における肢体運動の学習パターンには、歩行環境の変化後即時的に調節が行われる reactive feedback と、緩やかに調節が行われる predictive feedforward、という二つの調節様式があることが Reisman らによって報告された [12]。Reactive feedback では立脚時間やストライド長といった一側下肢の運動の結果発現される肢体内変数が左右下肢で非対称に適応し、ベルト速度が左右対称に戻ると後効果を引きずることなく元に戻る。一方、predictive feedforward では両脚支持時間とステップ長の両下肢の相互作用によって構成される肢体間変数が、はじめの左右下肢での非対称性が漸減していき、最終的に左右対称になる。その後ベルト速度を左右対称に戻すと、非対称歩行環境時の運動パターンの学習効果が残存しているために、しばらくぎこちない歩き方になってしまう。この現象は肢体間協調性の影響によるものとされている [12, 19]。こうした学習の形態に対して特定の時空間変数の対称､非対称に関する特徴的変化を追うことで、歩行の実行系と調節系を記述する。

これらを基盤に､Bastian らのグループと Duysens らのグループは研究対象を小脳疾患者にして、reactive feedback には脊髄レベルが、

predictive feedforward には小脳がそれぞれ関与することを示した[20-22]。つまり、小脳病変の対象者には歩行適応が生じず、後効果も現れなかった。これにより、split-belt 手法により捉えられる現象が歩行の神経制御系を反映したものであることをさらに支持した。

Split-belt 歩行の達成に関与する神経領域に病変がなく、歩行障害を呈する代表的な疾患に脳卒中がある。脳卒中は大脳を病変部位とし、片麻痺は非対称な肢体運動を生じさせ、立脚や遊脚時間[23]、両脚支持時間[24]、関節パワー[25]、関節可動域[26]そして歩幅[27]といった時空間変数の非対称性として現れる。Reisman らのグループは脳卒中片麻痺の歩行介入のツールとしての有用性を同分野の中で精力的に検証してきた。彼女らは左右非対称な歩行を呈する片麻痺患者に対して split-belt を利用して左右対称な歩行に修正することで、従来の歩行非対称性が軽減することや、その学習の影響が平地歩行などに汎化されることを報告した[28-31]。すなわち、これらの成果は神経学的異常から生じた歩行の左右非対称性はトレッドミルを用いた環境変化に伴い新たな歩行パターンを学習し、さらには後効果がその後の環境に汎化する学習効果を生むことを実証し、リハビリテーションへの可能性を広げた。

3.4. 解決課題

ただ、脳卒中片麻痺者に関するいくつかの研究では、片麻痺者の適応歩行時のステップ長は、健常者と同様に左右対称性を再獲得するに至ったわけではなかったことが触れられており、その要因については言及されていない[28-31]。片麻痺者にステップ長の非対称性がある場合、麻痺側に比べ非麻痺側を大きく出すタイプと、その逆のタイプがある[32, 33]。この元来の歩行の非対称性により、split-belt 課題を経て現れるステップ長

の対称性が異なった結果が現れる[34, 35]。片麻痺者における麻痺側と非麻痺側の差は、機能評価や介入効果のアウトカムにしばしば用いられる。同時に、リハビリテーション介入の目的となるが、こうした介入は非麻痺側を麻痺側に合わせるといった対応を招くことがあり、個人によっては望ましい戦略ではない可能性が指摘されている[32]。この個人間の結果の違いが何に起因するかを解決しなければ、split-belt treadmill の介入手段としての適用対象や設定条件を明らかにできず、リハビリテーションツールとして臨床応用の妨げとなる。

また、上記のような片麻痺のステップ長の結果に関する解釈を困難にする要因は他にも存在する。それは、健常人の split-belt 課題での歩行適応において、ステップ長が対称化する現象の合目的的な理由がわかっていない点である。ステップ長が定義されるのは踵接地の瞬間である。Split-belt 歩行中、両下肢の関節角度や両下肢に対する上半身の相対位置は、ベルト速度の遅い側（以下、遅側）下肢の接地時と、速い側（以下、速側）下肢が接地時では異なることが既知となっている[36, 37]。つまり、運動学データからの分析では左右下肢で非対称である点が多く、結果としてステップ長が対称化している意義は不明なままである。ステップ長のような時空間変数は運動の結果現れたものであり、それがベルト速度の非対称な歩行環境において、左右対称に統制する目的は運動の要因となる力学的分析をおこなわなければ知ることができない。歩行運動は身体重心と下肢からなる倒立振子運動としてしばしば扱われ、身体重心の並進運動にかかるエネルギーコストが最小化するように振舞っていると概念化されている。これを記述するためには、全身体を剛体ととらえ、その質点を代表点としてモデル化する剛体系モデルがある。Winter[38]は、質量比といった解剖学的データと重心位置といった力学的

データをもとにしたリンクセグメントモデルを提唱し、各セグメントを合算した CoM（Centre of Mass、身体質量中心）を扱った。しかし、split-belt 歩行の分析において CoM を推定した先行研究はない。さらには、CoM と床反力など下肢との力学的な関係性に言及したものもない。これは、先行研究におけるマーカートラッキングが、下肢のみ、または全身の最小限のランドマークで行われているために、CoM の推定が実験系的に困難であったことが要因の一つと予想される。

CoM は、正確な全セグメントをトラッキングした三次元座標データに、屍体を用いた調査に基づく身体の各セグメントの全身の質量に対する質量比、セグメント長、質量中心位置の情報を加えることで推定される。各セグメントの質量中心座標を取得し、大腿、下腿、足部の 3 セグメントから構成される下肢のように、複数のセグメントからなる系の質量中心座標を下記の計算により取得し、全身まで拡張する工程を経なければならない。以下、Xo=系の質量中心、M=系全体の質量、m=1 セグメントの質量、x=1 セグメントの質量中心。

$$Xo = \frac{m_1x_1 + m_2x_2 + m_3x_3}{M}$$

3.5. 研究目的と解決方法

本研究では、split-belt treadmill を用いて脳卒中片麻痺者の非対称歩行環境への歩行適応を調査する。Split-belt は 2 つのベルト速度差により、左右肢で異なる駆動速度でありながら、共同した駆動を実現しなければ適応することができない課題を提供する。脳卒中片麻痺者は片側の肢体運動が運動麻痺により障害されていながら、歩行適応に関する脊髄、小脳に病変を有さない者である。つまり、片麻痺者の歩行適応において

生じる肢体運動の変化は、歩行適応における肢体運動の果たす役割を明らかにし得る。

本研究の目的は、片麻痺者を対象にした調査によって、非対称歩行環境への歩行適応における肢体運動の影響を明確にすることである。そのために、まず片麻痺者の適応の違いを力学的に区別するための新たなパラメータの探求を行った。この検証では、適応歩行時にベルト速度の遅側、速側では、立脚期において下肢と全身の力学的な対応関係を表すパラメータに左右対称化が起きると仮説を立てた。これは、両下肢の相互作用から発現されるパラメータである点に加え、歩行適応が両下肢で力学的安定性が保ちエネルギーコストを最小化するよう収束していると予想したためであった。

次に、片麻痺者が持つ肢体運動における運動麻痺以外の非神経的要因の影響を考慮するため、関節への物理的制限が肢体間協調性に及ぼす影響の検証を行った。この検証においての仮説は、一側の一関節での単一の運動方向のみの制限であっても、肢体間協調性を低下させる影響があるとした。これは、肢体運動が各セグメントの質量分布や末端の軌道によってコーディネートされていると考えられ、それを物理的に非対称に操作されることは歩行適応を悪化させる影響をもたらすことが予想されたためであった。

そして最後に、片麻痺者での検証を行なった。本検証でも片麻痺者において麻痺側、非麻痺側下肢での相互作用から全身体の力学的安定を再獲得できる者には歩行適応が生じると仮説を立てて、検証を行った。

本研究は片麻痺者への split-belt treadmill を用いた介入の有用性を示すとともに、健常人の歩行適応時に見られる特徴的な時空間変数の変化が歩行制御に果たしている役割を明らかにし得る。

本研究では前述した課題に対し、これまで先行研究で明らかになった知見をもとに解決を図った。以下に項目別に論拠となる知見と解決方法を記載する。

3.5.1. 踵接地

ヒトの split-belt 歩行を対象にした先行研究によると、適応歩行では片側の歩行周期全体の時間的中心は当該下肢の単脚支持期の中心に一致するよう収束されることがわかっている[36]。歩行周期の始まりは当該下肢の接地、そして単脚支持期の開始は当該下肢の接地、終わりは反対下肢の接地により定義される。これはつまり、足の接地のタイミングがベルト速度の速側と遅側それぞれの歩行周期の位相中心を調整する働きをしている可能性を示している[36]。また、歩行中の猫の末梢神経電気刺激と運動ニューロンの発火イベントを記録した生理学的手法[39]やモーター付き関節と接地センサーを持った 4 足ロボットの各脚に、位相差を収束させる微分方程式を与えた位相振動子を適用したロボティクスによる実験[40]でも確認されている。これらは、歩行リズムとその位相を、感覚の求心性と摂動に基づく位相シフトおよびリズムの再設定（phase resetting）の生成により調節していることを示している。時空間変数の中でも時間変数は外乱になるような外的操作の影響を受けず、頑強な制御変数である[41, 42]。以上の知見を総合すると、踵接地のタイミングは、時間変数における規範を作ることで歩行適応に寄与していることが予想される。そこで、歩行周期においては踵接地のタイミングに着目して分析を行った。

3.5.2. 力学的安定性

歩行適応自体は周期的な歩行パターンを合目的的に変調させていることが予想される。つまり、一側ベルト速度の変化による摂動によって力学的に乱された歩行を、力学的に有利化するために肢体間協調性を利用してパターンを変更していると考えられる。歩行中の動的安定性を定量化するには、重心（CoM）および足圧中心（CoP; Centre of Pressure）の動きがどのように生成および制御されるかを理解する必要がある [43)]。転倒に関する先行研究は、踵接地した前脚を前方にスリップさせる課題で動的バランスを評価する際、CoM と前脚の CoP を解釈するには、それらを個別に調べるよりもより重要な示唆を与え得るとしている [44-45)]。中でも、踵接地タイミングは、両脚からの床反力が全身体に力学的な作用を与える両脚支持期の始まりである。先行研究では、両下肢の CoP と CoM の倒立振子モデル化を行い、歩行中踵接地直後の前脚 CoP と CoM のなす角度が前脚の制動成分の床反力ベクトルの角度と一致する条件が高い動的安定性と関与していたことを示した [46)]。これは歩行の開始、通常歩行、方向転換、歩行の終了それぞれでも一致しており、前脚の前方滑りを生じない安定した歩行の必須要素であろうと解釈されている。そこで、速側、遅側の踵接地タイミングでの力学的な安定性を、CoP と CoM の倒立振子モデルにより推定した。

3.5.3. 肢体の運動制限の影響の考慮

脳卒中片麻痺者には、運動麻痺による自動的な運動制限の他に、筋の痙性により他動的な運動制限が存在する。たとえば、ウェルニッケマン肢位に代表されるように肘関節の伸展が物理的に制限されているケースが多い。Split-belt treadmill での歩行適応には肢体間協調性が関与しており、肢体間協調性はセグメントの質量分布の変化によっても影響され

る程センシティブであることが既に示されている[47]。片麻痺者を対象にする場合、こうした物理的な関節運動制限による非神経学的な要因が結果に加味されていることになる。そこで、本研究では事前の可動域評価から股、膝、足、肘関節に可動域制限がある片麻痺者を除外して実施した。その他に、健常人に対して肘、膝関節それぞれに物理的伸展制限を付与した条件で肢体間協調性の変化と、力学的安定性への影響を調べた。これにより、片麻痺者の結果に対する神経的要因の解釈がより明確になることを目指した。

4. 研究の基本方法

4.1. Split-belt treadmill 実験プロトコル

被験者に対して、左右異なった速度制御が可能な 2 本のベルトで構成される床反力計内蔵ダブルベルトトレッドミル（1,000Hz、 Bertec 社製、 U.S.A.、 ITR5018-11）上での歩行を三次元動作解析装置 VICON（100Hz、 Vicon Motion Systems 社製、 UK、 MX T-series）で計測した後、全身に貼付した 39 個のマーカーの記録情報を元にソフトウエア Nexus2.8 にて剛体リンクモデル（Plug in Gait Full Body AI model）を構成し、運動学および運動力学データを抽出した（図 1)。歩行課題は両ベルト同様の速度で 3 分間歩行後、一側のベルトを加速度 0.5 m/s^2 で倍速にした非対称歩行課題を 6 分間継続させた。ベルト速度の変化について、そのタイミングと左右どちらのベルトが早くなるかに関する情報は被験者には伝えない。

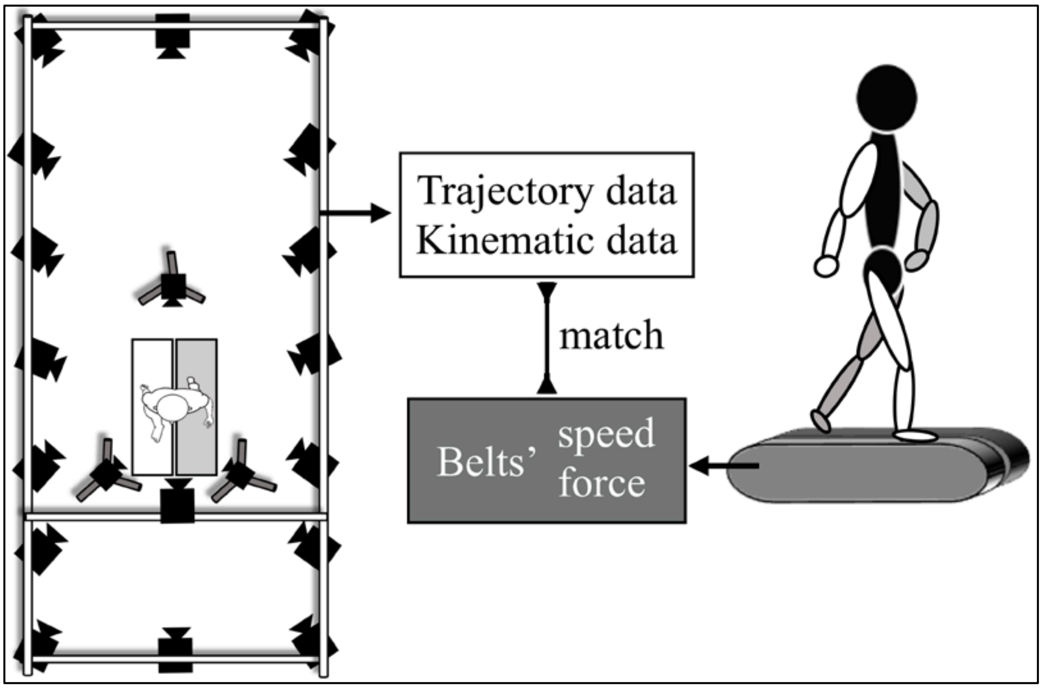


図 1. データ取得と同期

各相の定義は左右同速度歩行の最後の左右 5 歩ずつ、計 10 歩行周期を Baseline、ベルト速度非対称化直後の 10 歩行周期を Early-adaptation、ベルト速度非対称歩行の最後の 10 歩行周期を Late-adaptation とした（図 2)。トレッドミルの速度は LabVIEW（National Instruments 社製、 U.S.A.、 2016）にてプログラミング制御し、Vicon 内に速度情報を入力し、計測データと同期させることで相分けの正確性を担保した。

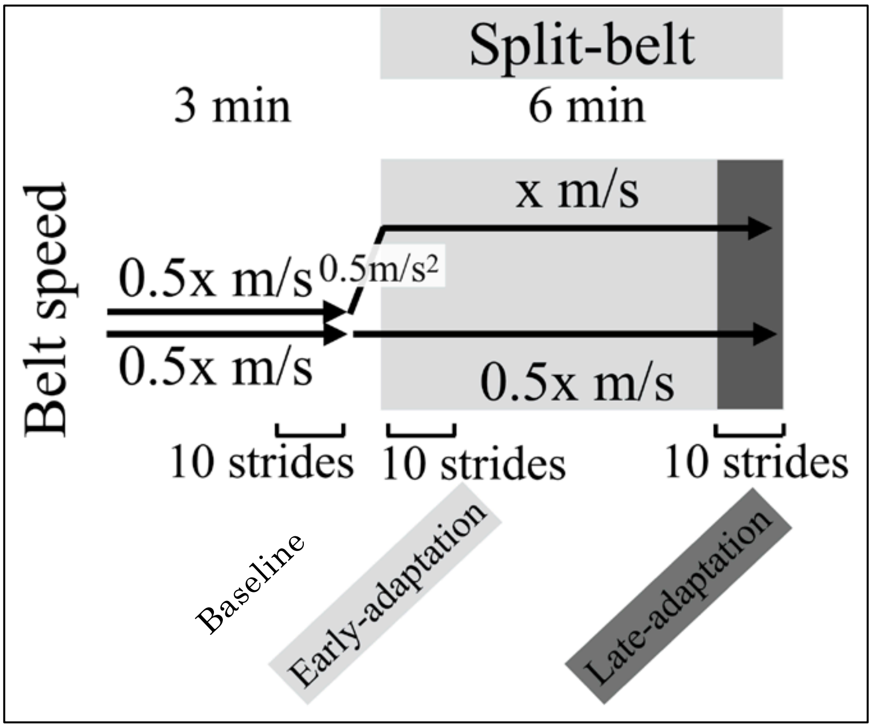


図 2. Split-belt treadmill 実験プロトコル

4.2. 変数

立脚時間、ステップ長、両脚支持時間を左右で図 3 のように定義し、基本的な時空間変数として扱う。一側肢によって定義される立脚時間は split-belt 歩行を継続しても左右非対称なままであるのに対し、両側肢によって定義される両脚支持時間とステップ長は split-belt 歩行により一時的に非対称化するものの、継続することで徐々に左右対称に適応するという特徴的な変化をする。これらの時空間変数の結果をもって、歩行適応がなされたかを論じる。

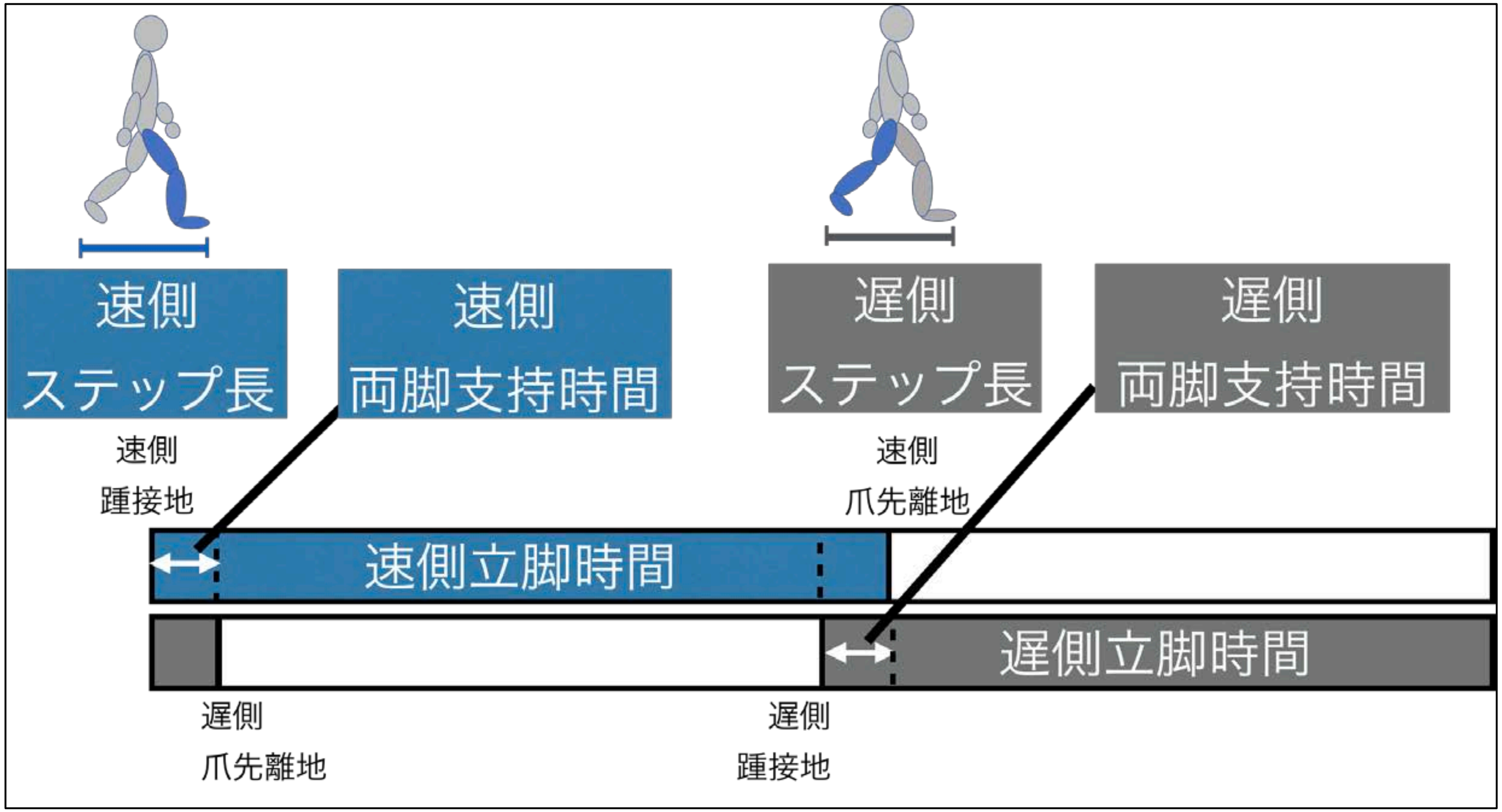


図 3. 左右の時空間変数の定義

4.3. 分析方法

計測した全ての生データに高周波数帯を減衰させるローパスフィルタである 2 次の Butterworth フィルタ（カットオフ周波数 5Hz）で波形処理を行った。歩行周期の定義には床反力鉛直成分を用いて踵接地、及び爪先離地を検出し、踵接地から爪先離地までの立脚期 100%時間正規化を行った。各相（Baseline、Early-adaptation、Late-adaptation）で抽出した右 5 歩分のデータを以下の symmetrical index の計算処理 [15] にて左右対称性を数値化した。

$$\frac{（ベルト速側 - ベルト遅側）}{（ベルト速側 + ベルト遅側）} \times 100\%$$

結果を従属変数、相（Baseline、Early-adaptation、Late-adaptation）を独立変数として、一要因の反復測定分散分析（repeated-ANOVA）を行い、多重比較検定（Bonferroni 法による補正）を行った。有意水準は 5%未満とした。

4.4. 倫理的配慮

本論文内の全研究はヘルシンキ宣言に則り、所属施設の研究倫理審査委員会の承認（承認番号 29501）を得た上、対象者に説明と同意を得て実施した。

5. 研究 I

5.1. 研究目的

主目的は Split-belt treadmill における歩行適応中に、ベルト速側と遅側の足接地時の動的安定性が再確立されることを実証することであった。

まず、CoM と CoP の相対的な位置関係が歩行適応によって速側と遅側で再確立されるかを確認した。接地した前脚の CoP と CoM を繋ぐ線と垂線のなす角度を変数として調査した。また、踵接地時の前脚末端と身体を表す他の変数の組み合わせについても歩行適応において対称性を再確立するもの、またその対称性の程度を比較検討した。

第二に、CoM と CoP からなる倒立振り子モデルを用いて、前脚の床反力制動成分のピークにおける後脚の影響を調べることで動的安定性を検証した。

5.2. 方法

5.2.1. 対象と計測

健常若年成人 15 名（平均 22±4.6 歳、男性 15）を対象に、上記した split-belt の基本方法をベースに速度は遅側 0.9m/s、速側 1.8m/s で統一して実施した（図 4）。

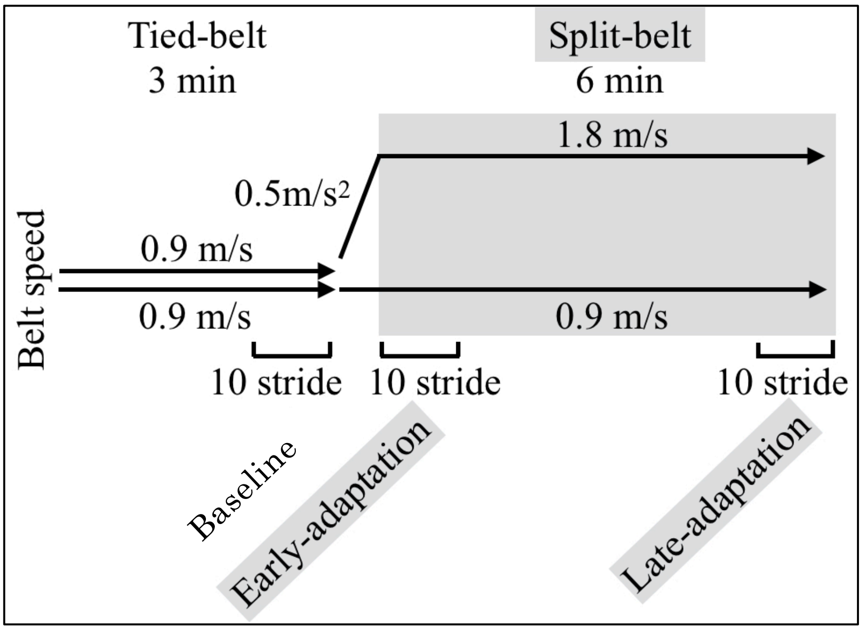
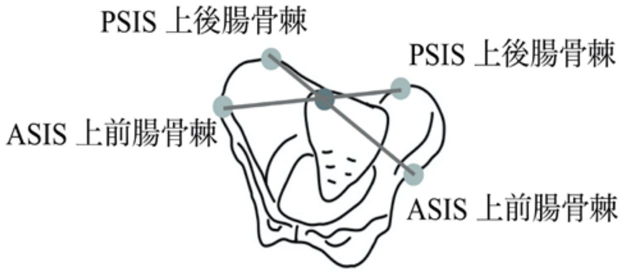


図 4. 研究 I の実験プロトコル

前脚末端側と身体中枢側を結んだベクトルが垂線となす矢状面角度を算出した。各変数は表 1 の通りで、全ての組み合わせで構成される角度は以下の通り。片側骨盤前後中心と外果（iPELANK)、片側骨盤前後中心と足圧中心（iPELCOP)、骨盤中心と外果（PELANK)、骨盤中心と足圧中心（PELCOP)、身体質量中心と外果（COMANK)、身体質量中心と足圧中心（COMCOP)。

表 1 前脚末端側と身体中枢側の変数の組み合わせ

中枢側		末端側
 片側骨盤前後中心（iPEL）		 外果（ANK）
 骨盤中心（PEL）	×	 足圧中心（CoP）
身体質量中心（CoM）		

5.2.2. 解析と統計分析

前脚の床反力垂直成分および前後成分から算出した前脚のピーク制動力は、通常、前脚における足接触の直後に生じる。これは、必要摩擦係数（RCOF）として知られており、歩行中のスリップに関連する重要な要素である。これまでの研究から、COMCOP の接線は、歩行中の RCOF と強く相関することが明らかにされている[48]。本研究では動的安定性を示すために、前脚の制動成分ピーク時の後脚の影響について分析

する。

まず、図 5 のように矢状面における CoM と CoP を用いた 2 足倒立振り子モデル化し、CoM 周りに働くモーメント（Mx）を表す式 1 を作成する。当式を前脚の制動力（Fy1/Fz1）が左辺に来るよう移項する。右辺は前脚の COMCOP の正接（tan θ 、式 3）と、それ以外の残差（RT、式 4）になるようまとめる。この残差の全ての項には後脚が関わっている。先行研究の結果は、接地直後の前脚の制動力ピーク時点で床反力ベクトルと前脚の COMCOP の接線が一致することがわかっている。したがって、前脚の制動力ピーク時点では、後脚に関する項（式 4）は 0 に限りなく近いことになる。

$$Mx = Fz1(y_{COP1} - y_{COM}) + Fz2(y_{COP2} - y_{COM}) - Fy1 * Z_{COM} + Fy2 * Z_{COM} \cdots \text{（式 1）}$$

$$\frac{Fy1}{Fz1} = \frac{y_{COP1} - y_{COM}}{Z_{COM}} + \frac{Fz2}{Fz1} * \frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} * \frac{Mx}{Fz1 * Z_{COM}} \cdots \text{（式 2）}$$

$$\tan\theta = \frac{y_{COP1} - y_{COM}}{Z_{COM}} \cdots \text{（式 3）}$$

$$RT = \frac{Fz2}{Fz1} * \frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} * \frac{Mx}{Fz1 * Z_{COM}} \cdots \text{（式 4）}$$

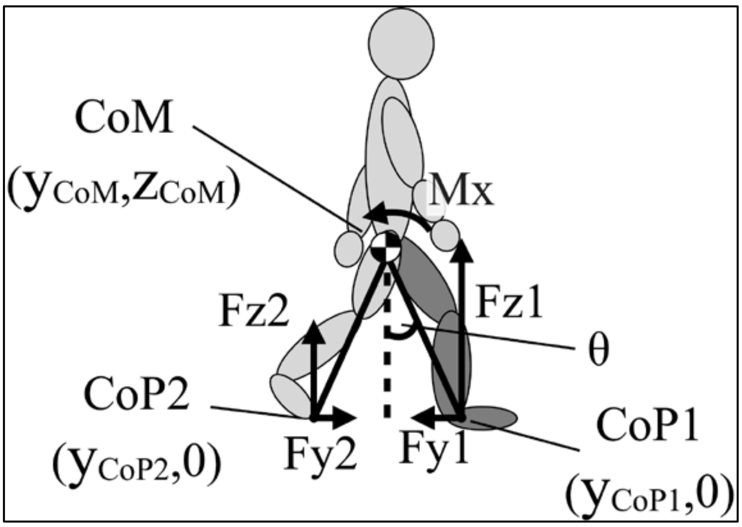


図 5. 二足倒立振子モデル

解析は末端側と末端側を結んだベクトルが矢状面上で垂直軸となす角度（仰角）を算出し、速側、遅側それぞれの踵接地時の角度を 5 歩ずつ抽出し、前述した index で対称性を数値化した。

全被験者の各相（Baseline、Early-adaptation、Late-adaptation）における対称性の平均を各相で 1 要因の反復分散分析を行った。また Late-adaptation での各変数に対して 1 要因の反復分散分析し、Bonferroni 補正を用いて事後検定を行った。

5.3. 結果

5.3.1. 時空間変数

図 6 に示すように、split-belt によってベルト速度非対称時には立脚時間は非対称なままで、Baseline との有意差があった。これに対し、両脚支持時間とステップ長は late-adaptation で Baseline との有意差がなくなり、左右対称に適応したのが見られるという一般的な結果であった。

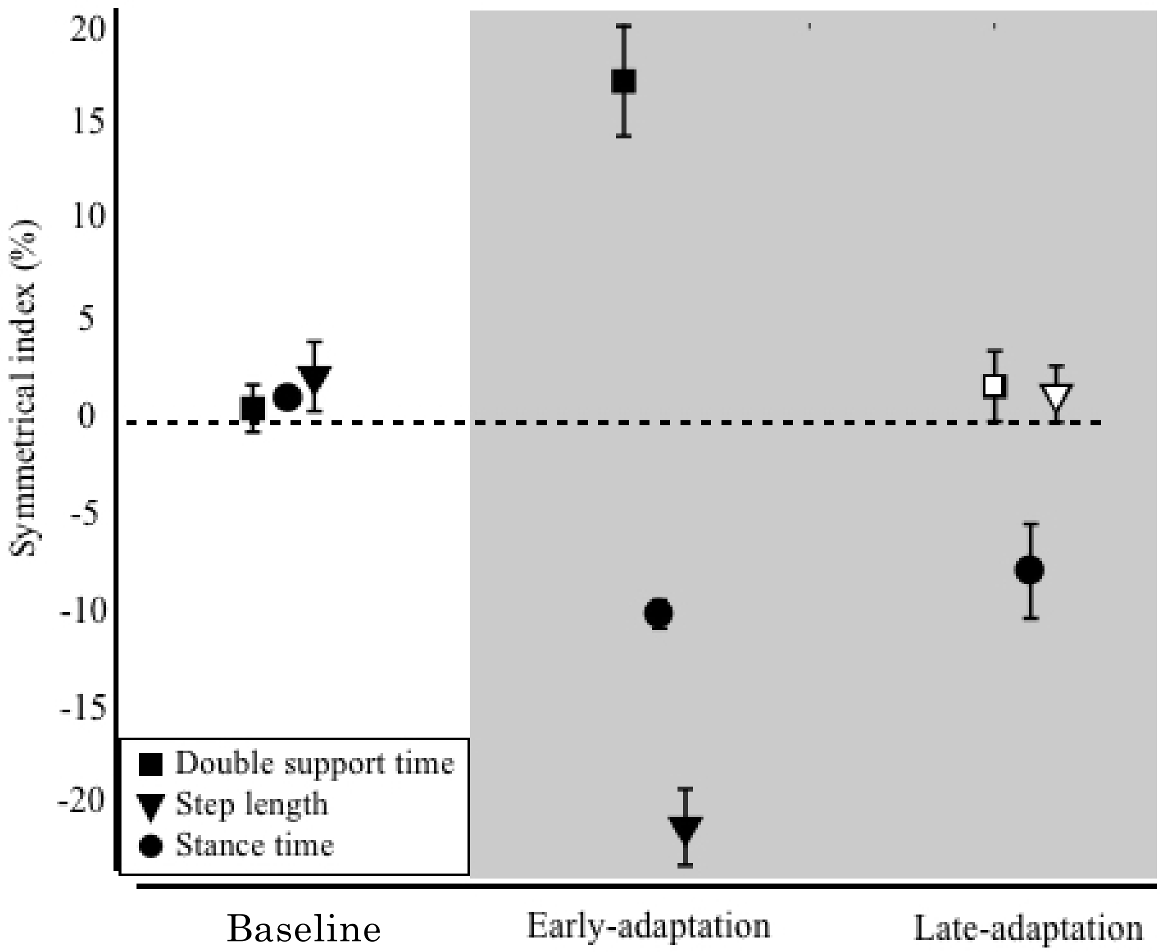


図 6. 時空間変数の結果

平均値と標準誤差のプロット。Early-adaptation および Late-adaptation では、Baseline と有意差ありの場合は黒塗り、有意差なしの場合は白抜き。

5.3.2. 前脚末端側と身体中枢側の変数

iPELANK、iPELCOP、PELCOP、COMANK は Late-adaptation でも Baseline と有意差があるか、Early-adaptation、Late-adaptation 共に有意差がない結果となった。PELANK と COMCOP は Early-adaptation で Baseline と有意差があり、Late-adaptation で有意差がなくなった（図 7)。

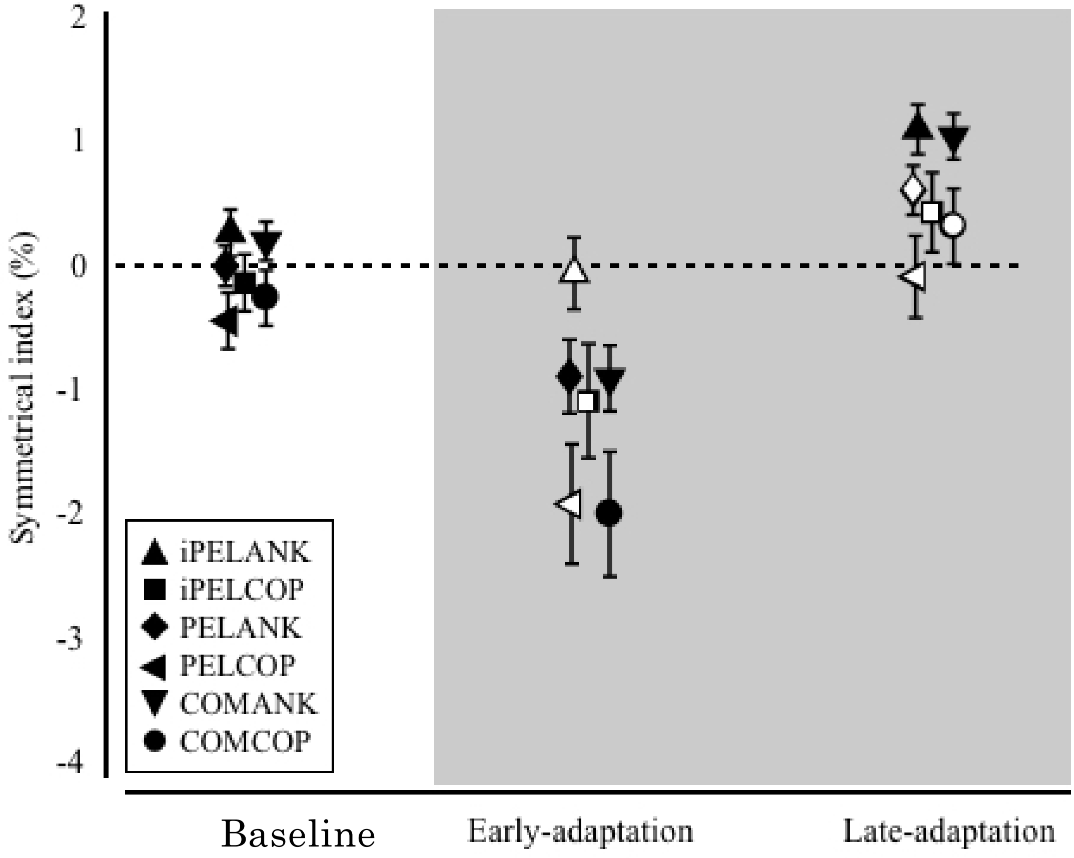


図 7. 変数の組み合わせの結果

平均値と標準誤差のプロット。Early-adaptation および Late-adaptation では、Baseline と有意差ありの場合は黒塗り、有意差なしの場合は白抜き。

Late-adaptation 内で変数の組み合わせ間を比較すると、PELCOP と iPELANK、COMANK が、COMCOP と iPELANK に有意差があった（図 8)。Early-adaptation で Baseline と有意差があり、Late-adaptation で Baseline との有意差がなくなった変数の中では COMCOP が最も Late-adaptation で対称に近似した適応をしていた。

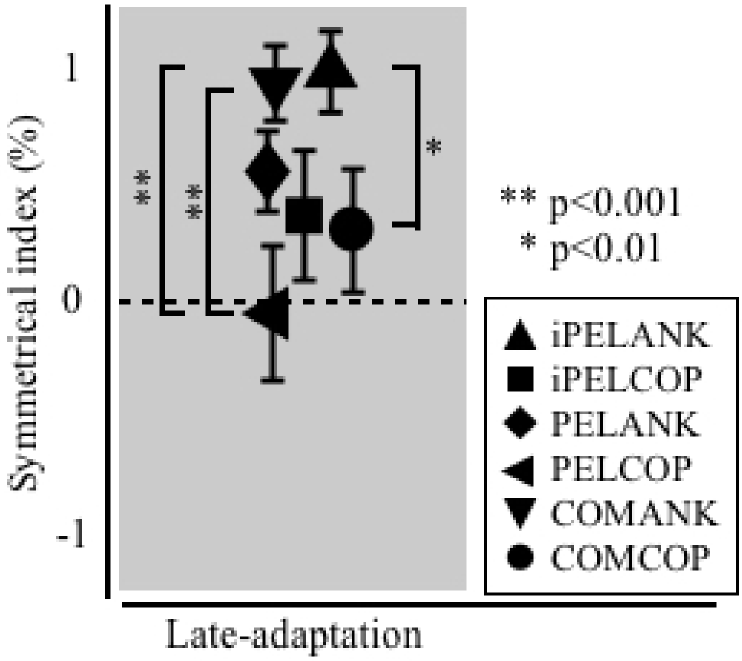


図 8. Late-adaptation での結果

平均値と標準誤差のプロット。*は標記の統計水準での有意差あり。

5.3.3. 力学的安定性（表 2）

前脚制動ピーク時の残差、残差の生じるタイミングの左右対称性、および前脚制動ピークタイミングは全てで Baseline と Late-adaptation に有意差がなかった。遅側の残差、残差のタイミングは Early-adaptation で一時的に Bseline と有意差が生じた。

表 2 力学的安定性に関する変数の結果

	Tied-belt Baseline		Split-belt Early-adaptation		Split-belt Late-adaptation	
	Fast	Slow	Fast	Slow	Fast	Slow
RT	0.09 ± 0.03	0.10 ± 0.04	0.08 ± 0.03	0.14 ± 0.04**	0.08 ± 0.03	0.10 ± 0.03
Symmetrical index of RT (%)	1.57 ± 12.72		30.42 ± 12.08**		13.74 ± 19.15	
RCOF timing (%)	12.15 ± 2.39	12.78 ± 2.17	12.42 ± 1.41	12.56 ± 2.60	11.68 ± 1.78	12.79 ± 2.97

** $p < 0.01$

RT は残差。Symmetrical index of RT は残差の左右対称性。RCOF timing は残差の算出に用いている前脚の制動成分がピークに達する相（立脚期 100%として正規化）。

5.4. 考察

本研究では split-belt treadmill 歩行時の適応において、ベルト速側と遅側の足接地時の動的安定性が保たれていることを実証するため、CoM-CoP 角とそれに類する前脚の末端側変数と身体中枢側変数の組み合わせのうち、split-belt 歩行中に左右対称性を再確立する変数の組み合わせを検討した。結果として CoM と CoP の組み合わせが高い対称性を確立する変数であることがわかった。

Split-belt 研究において、身体中枢側と接地前脚末端側をつないだベクトルと垂直線のなす矢状面上の仰角の変数としては、これまで Malone ら[42]が大転子と外果マーカーのベクトルのなす角度で定義した limb angle が用いられてきた。Malone らによって、ベルト速度が非対称であれば limb angle も非対称なままで適応変化しない変数であることが明らかになっており、本研究において類似した iPELANK が対称性を再確立しなかったことと一致する。Limb angle をはじめ、歩行適応における一側肢のみを表す変数は左右対称化する適応効果がないことは既知であり、本研究で用いた身体部位を変数とした他の組み合わせに適応効果がみられなかったことは先行研究に沿った結果であると言える。しかし、CoM-CoP 角のみは時空間変数において適応効果があった両脚支持時間とステップ長と同様に左右で対称に適応する変化をしていた。これは先行研究の結果を踏まえると、CoM-CoP 角が一側下肢のみを表す変数ではなく、両側肢の相互作用から表現される変数ということになる。

そこで、本研究の第二目的では、CoM と CoP の倒立振子モデルによる動的安定性（Dynamic stability）の検証を行なった。その結果、前脚の制動力がピークに達するタイミングで、後脚の推進力が CoM 周り

のモーメントに働く力は、速側、遅側共に通常歩行と有意差がなくなった。これは山口ら[48]によると、CoMと前脚CoPをつないだベクトル（tanθ）と、前脚の床反力の制動成分と垂直成分のベクトル（Fy1/Fz1）が一致し、前脚からの力のみがCoM周りに作用していたことになり、動的安定性の高さを示している。本結果も少なくとも左右ベルトが同一速度であるBaselineと有意な差がない動的安定性が速側、遅側ベルト上の下肢においても確保されていたことになる。

これらの結果は、健常成人ではsplit-belt歩行における適応に伴い、動的安定性が再確立されていることを示している。Reismanら[12]はsplit-belt歩行において踵接地後の体重移動時に前脚の角度を調整することは、CoMと前脚の相対位置を制御するために重要であると主張しており、それには力学的な安定性を再確立する合目的的な背景があることが本研究によって示された。また、踵接地直後の前脚の制動力ピーク時の前脚のCoM-CoP角は、後脚の推進による影響が0に限りになく近いために、両脚の相互作用によって現れている変数であることを示唆している。つまり、CoM-CoP角はステップ長や両脚支持時間と同様に、Bastianらのグループが定義したsplit-belt歩行時に左右対称に適応するinterlimb parameterであると言うことができる[12]。つまり、本研究によって接地直後の前脚CoM-CoP角は肢体間協調性（interlimb coordination）の寄与により予測的に制御される肢体間変数（interlimb parameter）と言うことができる。そして、split-belt treadmill歩行の適応において、Baselineのように左右の速度が同様な通常の快適歩行と同様の力学的安定性が再確立していることを示す変数として有用であると推察できる。

6. 研究Ⅱ

6.1. 研究目的

歩行における上肢と下肢の運動には空間的、時間的協調関係がある。Split-belt treadmill 課題はこの肢体間協調性を明らかにする手法の一つで、いくつかの先行研究で示されてきた。肢体間協調性はいくつかの神経障害をもつ対象において障害されていることが知られている。しかし、多くの有病対象者は関節運動範囲が制限されているなど非神経的要因のために空間的に非対称な身体を有している。よって、脳卒中片麻痺のような片側上下肢に運動麻痺と関節制限を持つ中枢神経系疾患者において split-belt により肢体間協調性の低下を認めた場合、神経的要因または非神経的要因によるものかが判別できない。

そこで、本研究では健常人の上肢、下肢それぞれに関節運動制限を付与した条件で split-belt treadmill における歩行適応動態を調査した。これにより、非神経的要因に起因する肢体運動の空間的非対称性が歩行適応に与える影響を明らかにすることを目的にした。また、歩行適応中におけるベルト速側と遅側の足接地時の動的安定性の再確立について検証した。

6.2 方法

6.2.1. 対象

健常若年成人 10 名（男性 10、肘制限条件 平均 22±1 歳、膝制限条件 平均 27±4 歳）を対象に、上記した split-belt の基本方法をベースに速度は遅側 0.9m/s、速側 1.8m/s で統一して実施した（図 4)。

7.2.2. 関節制限条件

肘関節運動制限は、片側肘関節を非伸縮性テーピング（CB-25, NITREAT）により伸展−20 度で固定した。屈曲に制限が出ないよう考慮した。膝関節運動制限は、片側膝関節を膝装具（REAQER）により伸展−20 度で固定した。屈曲に制限はないよう設定した。制限側は各個人でランダムに付与した。装具の質量を考慮し、反対側膝にも制限のない状態で装着をした。また、装具なしの条件に関しても、両側膝に制限のない状態での装具を装着して行なった。

Split-belt treadmill 課題としては、制限なし条件、制限側ベルトが高速化する速側肘、膝制限条件、制限と反対側ベルトが高速化する遅側肘、膝制限条件の 3 条件を実施した。制限なし条件をはじめに実施し、その他 2 条件の試行順序は各個人でランダムに行った。

図 9. 関節制限

6.2.3. 分析方法

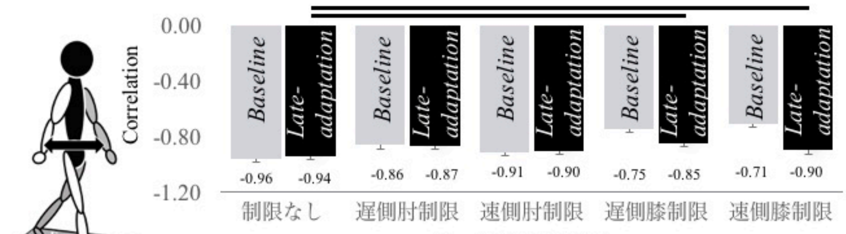
上下肢の肢体運動の変数は、上肢の腕振り、下肢の駆動が主に生じる矢状面上の振幅とした。上肢は肩峰に対する第 2 中手骨頭を、下肢は骨盤上前上後腸骨棘中心に対する外果の軌跡の振幅を抽出し、上肢と下肢で相互相関係数を算出することで、肢体間の逆位相、同位相での同期性を分析した。各条件と制限なし条件間の差を対応あり、各条件内の Baseline と Late-adaptaiton 間の差を対応なしの t 検定で分析した。

また、通常の時空間変数の対称性の分析に加え、足接地位置の対称性の変数に関しては、研究 I で用いた接地時の CoM-CoP 角とした。全被験者の各相（Baseline、Early-adaptation、Late-adaptation）における対称性の平均を各相で 1 要因の反復分散分析を行った。Bonferroni 補正を用いて事後検定を行った。

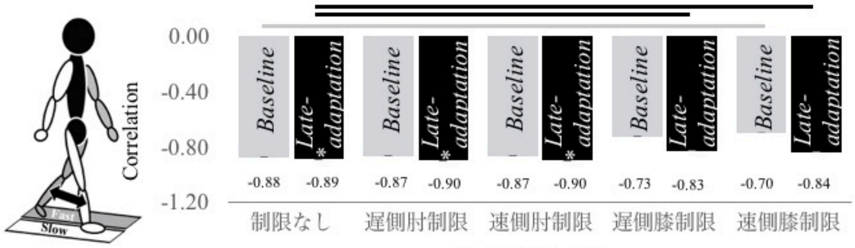
6.3 結果

6.3.1. 肢体間の相互相関係数

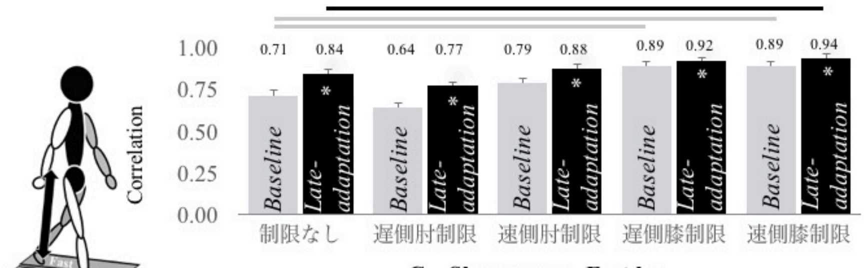
制限なし、および肘、膝関節制限の各条件の結果を図 10 に示す。肘関節制限条件では、制限なし条件に対し、速側肘制限条件での速側上肢と遅側下肢、速側上肢と速側下肢に相互相関係数の有意な差があった。いずれも Baseline に対して低下した。膝関節制限条件では、制限なし条件に対して有意な低下を示したのは、両上肢、両下肢であった。速側上肢と遅側下肢、もしくはその逆の組み合わせではむしろ有意な増大を示した。



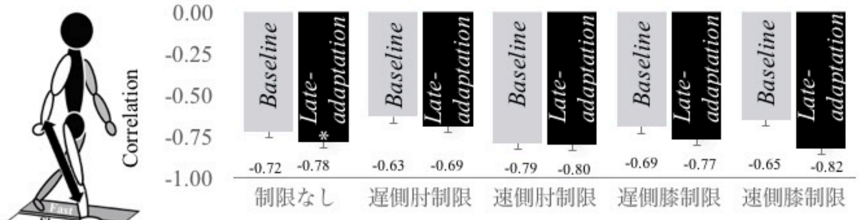
A　Arm vs. Arm



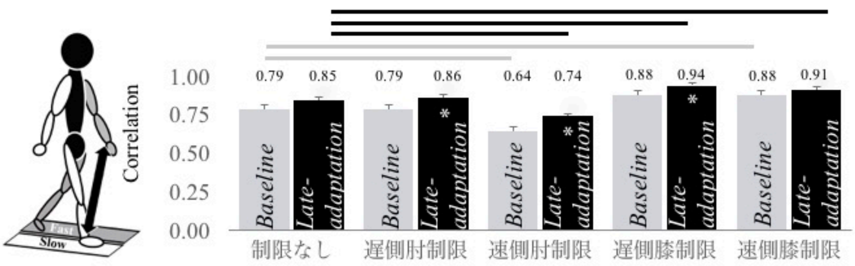
B　Leg vs. Leg



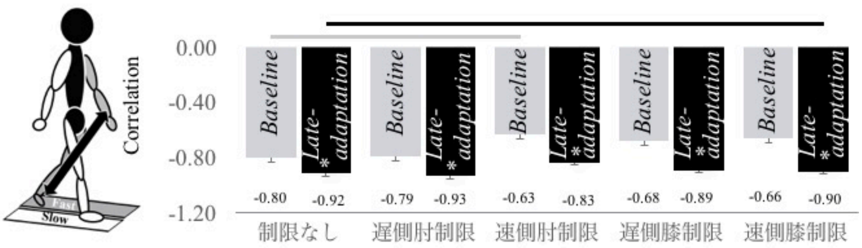
C　Slow arm vs. Fast leg



D　Slow arm vs. Slow leg



E　Fast arm vs. Slow leg



F　Fast arm vs. Fast leg

図 10. 各条件での相互相関係数の結果

水平バーは、制限なし条件とその他制限条件の Baseline（黒)、または Late-adaptation（グレー）同士で有意差あり（$p < 0.05$）を示す。

*は各条件の Baseline と Late-adaptation に有意差あり（$p < 0.05$）を示す。

6.3.2. 時空間変数と接地時の足部接地位置の変化（図 11）

全ての条件において、立脚時間は Early-adaptation と Late-adaptation が Baseline と有意差があった。それに対し、ステップ長と両脚支持時間は、Early-adaptation で有意差が出るが、Late-adaptation で Baseline と有意差がなくなる結果となった。これと同様に、CoM-CoP 角は Late-adaptation で Baseline と有意差がなくなった（図 12)。

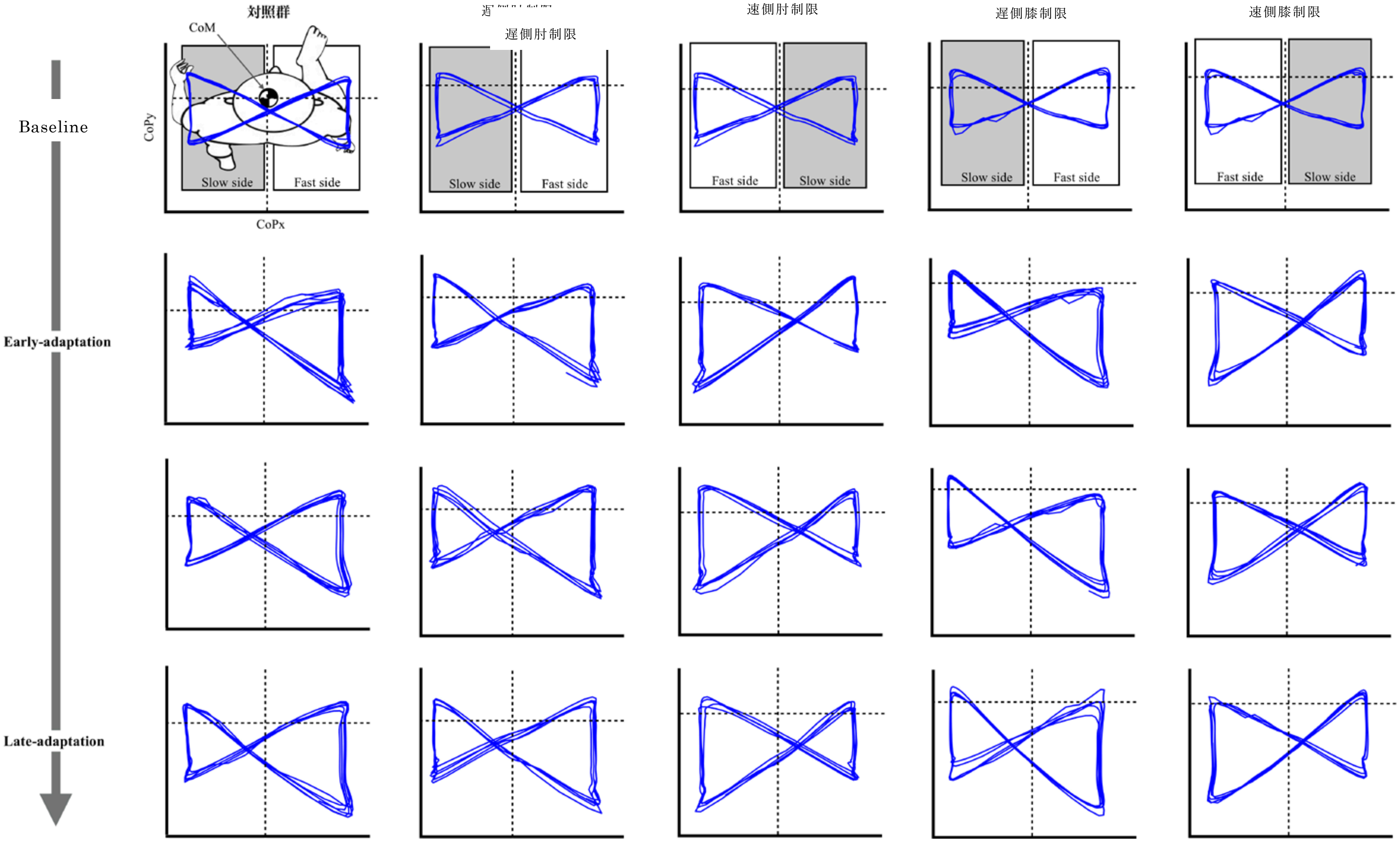


図 11. 条件ごとの時間に伴う CoP 軌跡の変化

CoM を原点とした左右合成 CoP の軌跡。

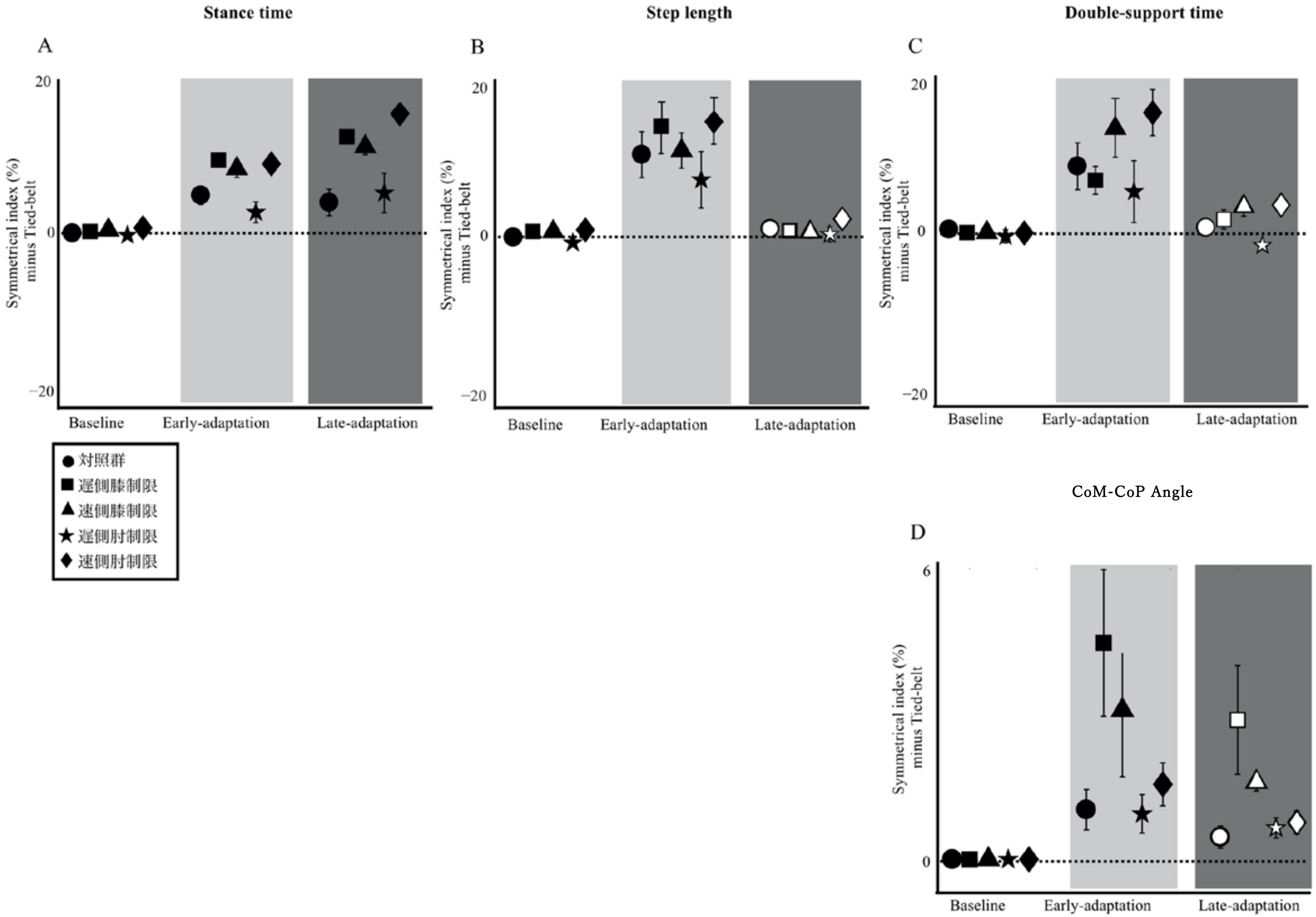


図 12. 時空間変数と CoM-CoP 角の結果

平均値と標準誤差のプロット。Early-adaptation および Late-adaptation では、Baseline と有意差ありの場合は黒塗り、有意差なしの場合は白抜き。

6.4. 考察

本研究では関節運動制限が片側上下肢に生じた際の split-belt treadmill 歩行における適応動態を検討した。結果として、一部の上肢と下肢の組み合わせにおいて肢体間協調性が低下していた。その一方で、時空間変数の適応と、全身に対する足接地位置に関しては関節運動制限のない条件と同様の適応効果が認められた。

肢体間の協調は通常、上下肢間の同側の神経接続を介して発現される[49)]。MacLellan ら[19)]は、中枢神経系の寄与に基づいて、split-belt treadmill 歩行中に上肢と下肢の前後振動の時間的連結が高まることを報告した。また、Bondi らは片側上肢に重りを負荷した条件で split-belt 歩行を実施した。その結果もまた、上下肢間での時間的連動性が低下し、脊髄を介した上肢と下肢の神経的連関による影響と結論づけた[47)]。実際に、脳卒中後片麻痺者やパーキンソン病者は、本研究で検討した拘束条件と同様に、上腕二頭筋やハムストリングスの痙縮や強直のために肘や膝の関節拘縮を呈しているケースが散見される。このような状態において、非神経的要因として、一側の一関節における一方向に生じた部分的な関節制限は、肢体間協調性を阻害することが明らかになった。

しかしながら、適応効果がある時空間変数（ステップ長、両脚支持時間）の結果から非対称歩行環境に対する歩行適応の観点では関節制限の影響はなかったと結論づけられた。さらには著者らが着目した足接地位置の左右対称性の再確立に関しても、関節制限の影響は認めなかった。特に、膝関節伸展制限条件は踵接地時の下肢長の延長に直接的に影響する可能性が考えられた。しかし結果としては、制限側と非制限側のそれぞれの下肢と全身体との相対位置は、同様になるよう調整がなされていることが推察された。

これらから、少なくとも健常人においては、split-belt 歩行課題により強化される肢体間協調性が、関節運動制限によって阻害されるが、肢体間の相互作用によって生じる時空間変数と、下肢接地時の位置の適応には影響を及ぼさない調節が可能であることがわかった。それと同時に、下肢接地位置の左右対称性の再確立は、時空間変数の対称性の再確立と強く連動しており、歩行適応の効果を示す変数として有用であることを示唆していた。

ただし、遅側膝制限条件は他の条件と比べて Late-adaptation での標準誤差が大きく、差がない可能性平均値の結果として統計的な差を認めなかったとの結論に過ぎず、解釈は慎重にすべきである。

7. 研究Ⅲ

7.1. 研究目的

本研究の目的は split-belt treadmill における歩行適応課題において、片側上下肢に運動麻痺を生じた脳卒中片麻痺者の適応過程を検証することであった。同様の対象者で行われた split-belt 実験の結果は、適応効果に個人差があり、その個人特異性の要因が不明であった。本研究では適応効果の指標として、ステップ長と両脚支持時間の他に、研究Ⅰ、Ⅱで検証した足接地位置の対称性を用いて、検討を行った。

7.2. 方法

7.2.1. 対象と計測環境

発症後 6 ヶ月以上経過した慢性期脳卒中片麻痺者 22 名（男性 17、平均 67±9 歳）及び、年齢と性別比をマッチングさせた健常高齢者 9 名を対象とした（表 3)。除外基準は他の神経的、整形外科的既往、コントロール不能な高血圧、ペースメーカー使用、タスクの遂行ができない状態とした。片麻痺者には下肢の麻痺の重症度を FMA（Lower Furgl-Meyer Assessment)、日常生活自立度を FIM（Functional Independence Measure)、歩行機能を TUG（Timed Up and Go Test）で評価した。

表 3. 基本情報

Subject	Age	Sex	Hemiparetic side	Months since stroke	LE FMA (/34)	FIM	TUG (s)	Speed slow/fast
S1	68	M	R	128	21	115	17.80	0.25/0.5
S2	63	F	L	27	20	115	31.10	0.3/0.6
S3	69	M	R	142	29	126	7.00	0.45/0.9
S4	75	M	R	120	22	115	15.80	0.35/0.7
S5	70	F	R	65	21	120	14.00	0.3/0.6
S6	79	M	L	13	31	105	12.00	0.2/0.4
S7	73	M	L	36	26	120	16.00	0.25/0.5
S8	78	M	R	9	34	126	10.40	0.4/0.8
S9	65	M	L	7	25	120	13.97	0.35/0.7
S10	71	M	R	18	20	114	18.80	0.2/0.4
S11	73	M	R	8	22	107	25.90	0.2/0.4
S12	70	M	L	141	19	104	45.12	0.2/0.4
S13	57	M	L	18	19	118	30.30	0.2/0.4
S14	45	F	L	66	21	114	15.80	0.2/0.4
S15	48	F	L	132	14	119	28.59	0.2/0.4
S16	84	M	R	7	29	121	24.80	0.2/0.4
S17	71	M	R	162	27	109	21.81	0.2/0.4
S18	70	M	R	33	20	106	32.50	0.2/0.4
S19	63	M	L	102	28	83	46.88	0.2/0.4
S20	65	F	L	47	28	121	21.82	0.2/0.4
S21	69	M	R	50	25	102	27.00	0.2/0.4
S22	55	M	L	103	14	116	37.65	0.2/0.4

片麻痺被験者の基本情報。経過期間（Months since stroke）、下肢 FMA（LE FMA）、ベルト速度（Speed）。

片麻痺者に関しては、屋外歩行時に短下肢装具を装着している場合は着用を許可し、図 13 のように転倒防止用のハーネスを着用した。ハーネスが免荷にならないよう留意した。Split-belt treadmill 課題の速度設定は図 2 のようにした。本試行に先駆けて測ったトレッドミルでの最大歩行速度を x とし、速度比は 2:1 とした。健常高齢者はこれまでの研究Ⅰ、Ⅱと同様の 0.9、1.8 m/s の速度比 2:1 で行った。

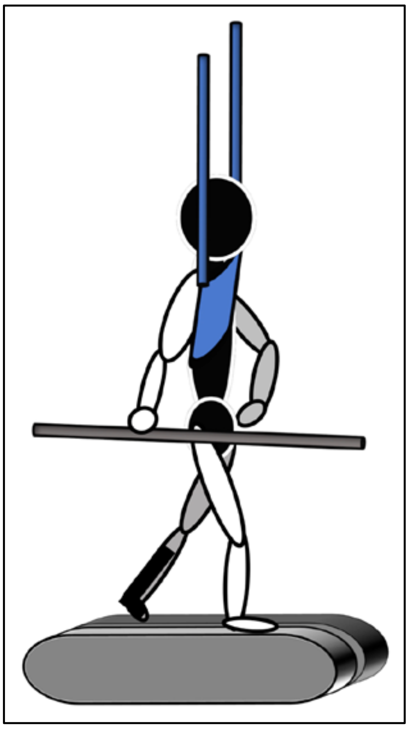


図 13. 片麻痺者の split-belt treadmill 歩行時の環境

7.2.2. 分析方法

片麻痺者において、Late-adaptation で CoM-CoP 角の対称性を再確立したグループを"responder"、しなかったグループを"non-responder"にサブグループ化した。

時空間変数の対称性を、3 つ相（Baseline、Early-adaptation、Late-adaptation）と 2 つのサブグループの 2 要因の反復測定分散分析にて、交互作用が認められた場合、多重比較を行った。CoM-CoP 角の対称性はサブグループ間で対応のない t 検定で比較した。

また、研究 I で用いた矢状面における CoM と CoP の 2 足倒立振り子モデル化を用いた分析を実施した。具体的には、式 2 の前脚の制動力（Fy1/Fz1）のピーク値 RCOF と、式 3 の前脚の CoM-CoP 角の正接である tan θ との有意差をグループ内で対応のある t 検定で比較した。

$$Mx = Fz1(y_{COP1} - y_{COM}) + Fz2(y_{COP2} - y_{COM}) - Fy1 * Z_{COM} + Fy2 * Z_{COM} \cdots (式\ 1)$$

$$\frac{Fy1}{Fz1} = \frac{y_{COP1} - y_{COM}}{Z_{COM}} + \frac{Fz2}{Fz1} * \frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} * \frac{Mx}{Fz1 * Z_{COM}} \cdot\cdot\cdot (式\ 2)$$

$$tan\,\theta = \frac{y_{COP1} - y_{COM}}{Z_{COM}} \cdot\cdot\cdot (式\ 3)$$

7.3. 結果

7.3.1. サブグループ

健常高齢者において Late-adaptation での CoM-CoP 角の対称性は平均 0.78 であったため、この値を片麻痺者のサブグループ化の閾値に設定した。サブグループの結果を表 4 に示す。t 検定の結果 CoM-CoP 角を除き、2 グループ間に有意差はなかった。

表 4 各条件のサブグループの結果

	Paretic leg slow			Paretic leg fast		
	Responder (n = 8)	Non-responder (n = 14)	*p* value	Responder (n = 5)	Non-responder (n = 17)	*p* value
LE FMA (/34)	22.2 (7.0)	23.8 (4.6)	0.52	21.2 (7.1)	23.9 (5.0)	0.35
FIM (/124)	114.7 (7.8)	112.7 (10.8)	0.64	111.4 (5.0)	114.0 (10.7)	0.60
TUG (s)	21.9 (10.7)	24.3 (11.1)	0.63	22.0 (10.1)	23.8 (11.2)	0.75
Maximum walking speed (m/s)	0.5 (0.2)	0.5 (0.1)	0.38	0.5 (0.1)	0.5 (0.1)	0.59
Symmetrical index of percent stance time during baseline period	7.9 (5.5)	6.3 (5.9)	0.54	8.4 (4.1)	7.0 (5.3)	0.49
Symmetrical index of step length during baseline period	32.7 (30.4)	36.3 (29.0)	0.70	50.9 (33.8)	31.7 (29.9)	0.44
Symmetrical index of percent double-support time during baseline period	18.9 (12.1)	13.6 (11.8)	0.16	16.2 (15.0)	15.1 (9.4)	0.18
Symmetrical index of CoM-CoP angle during late-adaptation period	0.28 (0.20)	2.43 (2.09)	<0.01	0.43 (0.21)	2.73 (2.19)	<0.05

カッコ内は標準偏差。

7.3.2. CoM-CoP 角

CoM-CoP 角の時系列変化（図 14A）では、健常高齢者と responder が Early-adaptation 以降に左右対称を示す基線に近づくのが観察された。Late-adaptation で CoM に対する相対的な CoP の軌跡でも non-responder は接地時点の最前方位置に左右差があることが観察される（図 14B)。

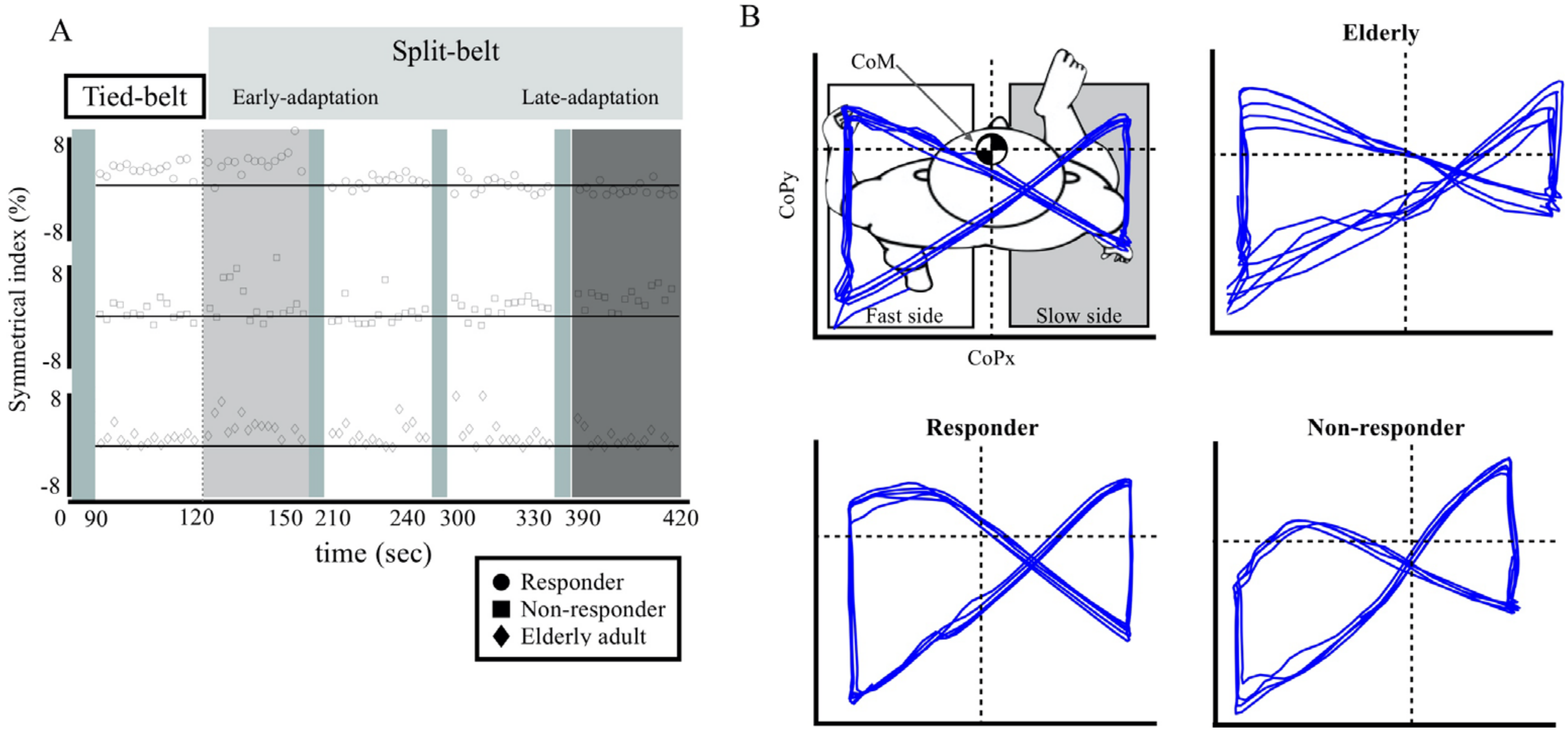


図 14. CoM-CoP 角の時系列変化と CoP の軌跡

7.3.3. 時空間変数

時空間変数の結果は、立脚時間では麻痺側のベルト速度が高速化する条件での responder を除き、全てが Late-adaptation で Baseline と有意差があり、非対称のままであった（図 15A、D)。ステップ長と両脚支持時間はベルト条件によらず non-responder は非対称なままであった。以下、F（自由度）＝F 値、p=有意確率の順に記載する。ステップ長は健常高齢者群と交互作用があり、麻痺側遅側条件では F（2、66）＝5.67、p＝0.01、麻痺側速側条件では F（2、72）＝3.73、p＝0.03 であった。両脚支持時間は健常高齢者群と交互作用があり、麻痺側遅側条件では F（2、66）＝2.40、p＝0.09、麻痺側速側条件では F（2、72）＝2.50、p＝0.09 であった。

それに対し、健常高齢者と responder は Late-adaptation と Baseline に有意差がなくなり、対称性を再確立した。健常高齢者群と交互作用はなし。ステップ長は麻痺側遅側条件で F（2、42）＝0.17、p＝0.85、麻痺側速側条件で F（2、36）＝0.15、p＝0.86 であった。両脚支持時間は麻痺側遅側条件で F（2、42）＝0.32、p＝0.73、麻痺側速側条件では F（2、36）＝0.36、p＝0.51 であった。

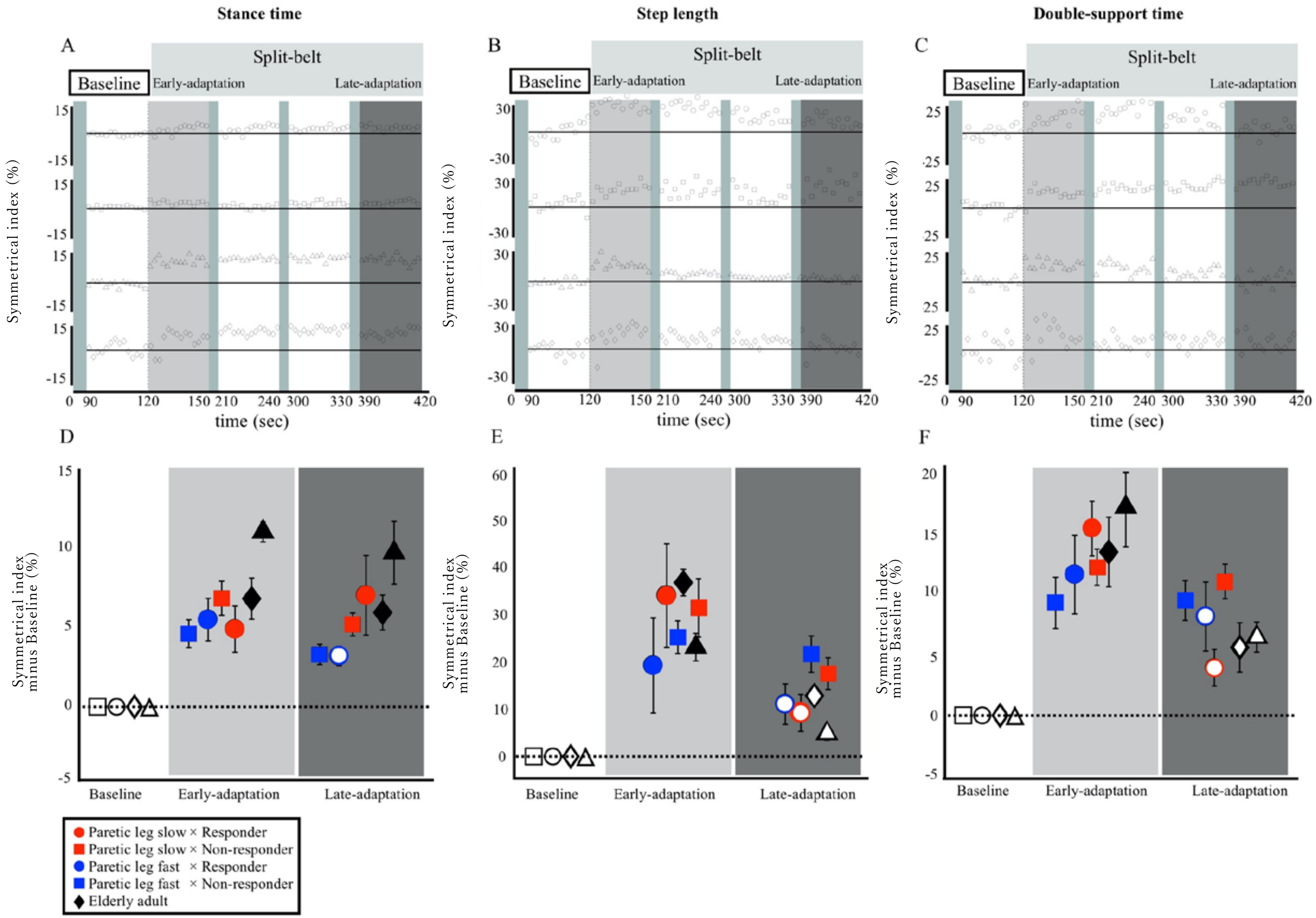


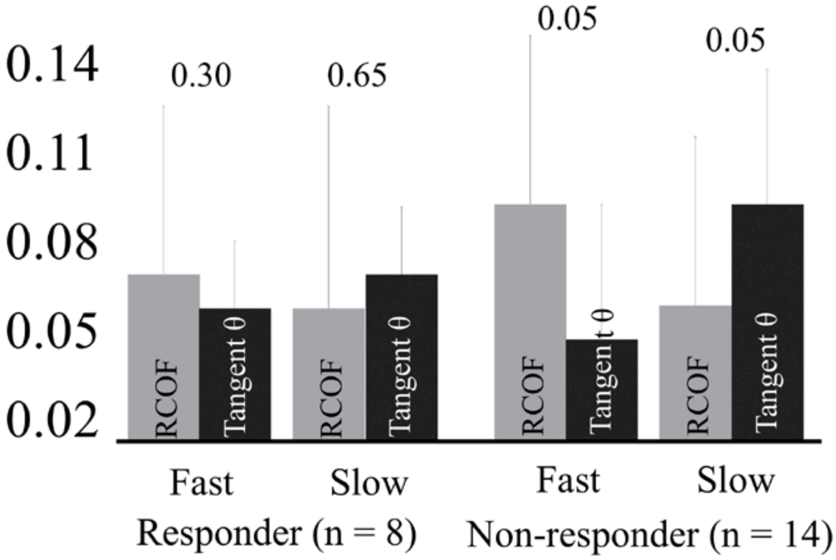
図 15. 時空間変数の結果

Early-adaptation および Late-adaptation では、Baseline と有意差ありの場合は塗り潰し、有意差なしの場合は白抜き。

7.3.4. 力学的安定性

Late-adaptation における Fy1 / Fz1 と tan θ には、いずれのベルト条件でも responder に有意差を認めなかった。一方、non-responder は麻痺側遅側条件では有意差は認めなかった（p = 0.05）が、麻痺側速側条件では有意差を認めた（p < 0.05）。

A



B

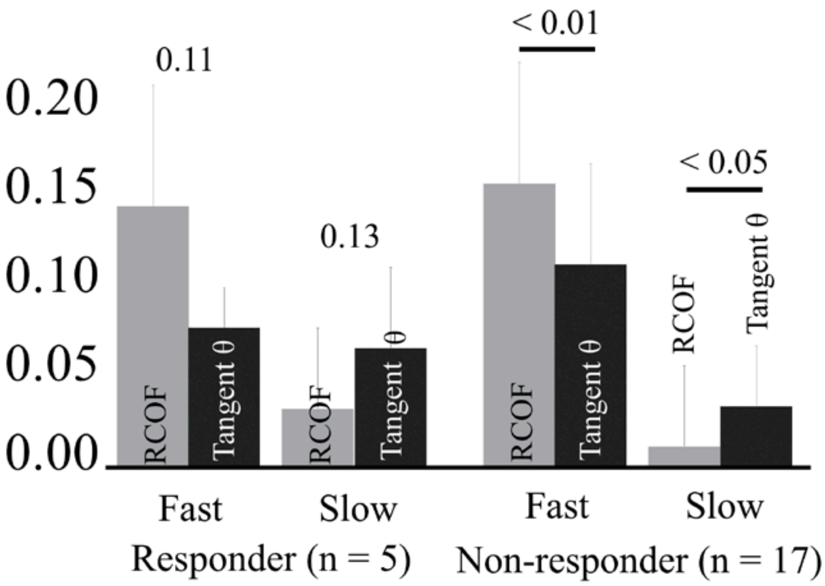


図 16. Fy1 / Fz1 と tan θ の差（A:麻痺側遅側、B:麻痺側速側）

7.4. 考察

研究Ⅲでは split-belt treadmill における歩行適応課題において、片麻痺者の適応過程を検証した。適応効果の有無によって対象者を分類するにあたり、研究Ⅰ、Ⅱで検証した CoM-CoP 角の対称性に再確立を指標として分析を行った。その結果、CoM-CoP 角が表す全身との相対的な足接地位置の対称性の再確立は、時空間変数における適応効果の患者特異性を区別する指標となっていた。

Late-adaptation で CoM-CoP 角の対称性を再確立した responder と健常高齢者は、共に Baseline と同等のステップ長と両脚支持時間の対称性を再確立していた。ただし、responder における時空間変数の対称性に関しては、その対象者の Baseline での対称性と同程度の対称性に Late-adaptation で収束したという結果であり、完全左右対称ではなかった点では先行研究[28, 29, 31]と同様であった。この結果は、歩行適応においては、CoM-CoP 角が表現する力学的安定性に関与する全身に対する踵接地位置は、左右ベルト速度が異なっていても左右で対称に収束させることが必要な条件であることを示している。また、片麻痺者にとって split-belt treadmill 歩行中の時空間変数は、完全な左右対称に再確立されることが歩行適応に必須な要素ではない可能性が示された。つまり、本研究の responder のように、ベルト速度が左右対称の通常歩行時の時空間変数の非対称性と同等に収束することが歩行適応においては必須な要素だと予想する。また、床反力[50)]、時間因子[36)]に関する先行研究の結果から、足の接地が split-belt 歩行の適応における重要な要素である可能性が示されている。これらの先行研究を踏まえ、片麻痺者の時空間変数の結果の違いは、足接地位置が重大な運動学的寄与をしていることが考えられる。

また、倒立振子モデルからの力学的分析により、前脚の CoM-CoP 角（tan θ）と制動成分のピークベクトル（Fy1/Fz1）に統計的な差を認めなかったことから、CoM-CoP 角の対称性の違いに関連していることが明らかになった。床反力の垂直成分と前後成分から計算される前脚の最大制動成分は、接地直後に生じる。これは踵と床間の摩擦係数を超えてはならないため、歩行中のスリップに関する重要な因子である[51)]。直進やターン時の前脚の最大制動力は、tan θ と強い正の相関がある[48, 52, 53)]。responder では、健常高齢者と同様に、tan θ と Fy1 / Fz1 に統計的有意差を認めなかったのに対し、non-responder は tan θ と Fy1 / Fz1 に有意差があった。Fy1 / Fz1 式を表す式 2 において、tan θ にあたる第 1 項を除く 2 項目以降は Fy2 や Fz2 が入るため後脚の影響を含んでいる。つまり、tan θ と Fy1 / Fz1 に差を認めない responder では、接地時の CoM-CoP 角と前脚の床反力ベクトルに差がないことになり、CoM 周りに作用する力が前脚の制動力のみとなっていることを示している。

逆に non-responder では後脚による推進力が CoM 周りに作用していることとなる。遅い歩行のような動的安定性が低下すると、tan θ と Fz1 / Fy1 の差は大きくなる[54)]。したがって、CoM-CoP 角の対称性を再確立できなかった片麻痺者は、split-belt 歩行での動的安定性が低かった可能性が示唆される。ただし、本研究では non-responder と responder でのサブグループ間には、臨床評価尺度および他の計測データの結果に有意差が認められなかった。以前の研究では脳卒中者において、動的安定性のスコアと動的バランス評価の高さには関連が見られず、同様の性質を示す所見に関しては、慎重に解釈すべきであると指摘されている[55-56)]。

8. 考察

8.1. 各研究結果の総合的解釈

本研究は片麻痺者の歩行適応動態を分析することで、非対称歩行環境への歩行適応における肢体運動の影響を明確にすることを主目的に行なった。そのために、歩行周期のうち踵接地期における全身体と接地下肢との力学的安定性を分析し、肢体に生じた物理的運動制限の影響を考慮した上で、片麻痺者における結果の患者特異性を区別することを一連の研究で試みた。

結果として、踵接地時に接地した前脚の CoP と CoM との矢状面上の相対位置を示す CoM-CoP 角は、健常人では厳格に左右で対称化するよう統制されていた。この CoM-CoP 角の統制は、他の変数に比べ再確立される対称性が高い上、肘や膝関節の物理的関節運動制限の影響を受ける可能性が低かった。そのため、split-belt を用いた歩行適応を表すパラメーターとしての有用性が確認された。最後に、片麻痺において、CoM-CoP 角の対称性が通常歩行と同様に再確立できたか否かをもとに分類したサブグループ間での検討では、従来の時空間変数の対称性に統計的有意差が認められた。よって、歩行環境変化への適応が可能か否かにおいては、肢体の運動制限や運動麻痺に関わらず、接地時の力学的安定性を再確立できることが条件である可能性が示された。つまり、肢体運動が歩行適応に果たしている役割は、接地した前脚からの制動力でのみ全身体の加速をコントロールするのに有利な下肢と全身の相対的な位置関係にする目的を達成することに貢献していると考えられた。

8.2. 2 足歩行の安定化機構

2 足歩行の力源について、アクティブなエネルギーコストは最小限に

抑え、受動運動によって主に生成されていると考えられている[57]。前脚が床面と接触すると身体に向かう方向の反力が発生する。これは矢状面において CoM を後方に回転させるモーメントとなる。Kuo らによる理論的倒立振子による動的歩行モデルでは、この負の仕事は、足が接触する直前の後脚での push off にて軽減されると言われている[58]。さらに Kuo らは、この前脚接地直前の後脚の push off は、予測的に安定した歩行を維持するために必要なアクティブな仕事量の軽減に作用することを示している[59]。上記した後脚の push off に関する知見は、本研究で示された踵接地時に後脚による推進の影響が無くなった結果と合致する。先行の split-belt treadmill 研究において、踵接地時の床反力制動成分と前脛骨筋活動には学習効果を認めることから、踵接地が予測的に制御される要素である可能性が示されている[50]。本研究によって、前脚を力学的有利な身体との相対位置にあらかじめ配置することに、接地直前までの後脚の push off が関与していると予想された。これには両下肢において予測的な相互作用関係が成り立っており、肢体間協調性の定義に即している。よって、非対称歩行環境での適応時の歩行は、肢体運動を用いた予測的な身体の安定化を図っている状態であることが考えられた。

8.3. 脳卒中片麻痺者に対する臨床的示唆

Reisman らによる 2005 年の研究では、CoM と前脚の相対位置を制御するために、体重移動時に前脚の角度を調整することの重要性が主張されている[12]。我々の研究では実際に CoM の推定を行い、この主張を支持する結果が導かれた。これに加えて本研究では、肢体運動に麻痺後遺症を有する片麻痺者においても、CoM と前脚の相対位置を左右の接地時で対称性を再確立することができた者は、split-belt 課題において健常

者と同様に歩行適応が可能であることを示した。前出の章 4.3.にて述べたように、脳卒中による大脳皮質病変を主とするため、split-belt 課題遂行にあたっての病態特異的な問題はない。今回、健常人と同様な歩行適応が実現されなかった片麻痺者には、歩行継続中に肢体もしくは体幹において、それぞれの相対位置を修正させられない要因があったことが考えられる。この相対位置の修正は、研究Ⅱから物理的運動制限だけでは妨げられないことを確認している。Split-belt treadmill での歩行適応は、入力情報に基づく出力で try and error を繰り返しながら、歩数を重ねるごとに誤差修正していく過程である[50)]。つまり、片麻痺者の場合、入力情報に誤りが生じるか、意図した出力と実際の出力に齟齬が生じているかが考えられる。片麻痺者の場合、ベルト速度変化による摂動後、一時的に乱された接地位置を、立脚期接地時に入力される下肢からの情報や、遊脚期に意図する接地位置を実現できないことが予想される。脳卒中が呈する症候の中では、荷重感覚といった深部感覚鈍麻によって接地時に入力される情報に誤りが生じ、誤差修正が行われないことが考えられる。

また出力に関しては、下肢側と体幹側、もしくは麻痺側と非麻痺側で双方向に修正していく必要があり、そこに問題を生じていた可能性も考えられる。つまり、麻痺脚の接地にあたり、前方配置を妨げる病態を呈した片麻痺者は split-belt treadmill による介入の適用になりにくい可能性がある。例を挙げると、遊脚時の膝伸展運動を妨げるハムストリングスの強い痙性や、足底に生じる表在感覚鈍麻、下肢の関節位置覚の感覚鈍麻といった状態などが想定される。このように、本研究は split-belt treadmill を片麻痺者へのリハビリテーションツールとして適用する際の知見を提供した。

さらに、今回歩行適応を達成した片麻痺者は、下肢接地位置に関しては麻痺側、非麻痺側で対称性を確立したが、時空間変数に関しては各個人の通常歩行時と同程度の対称性を再確立させたにとどまる結果であった。この時空間変数の結果は、章 8.2.でも言及したように、先行する片麻痺者に split-belt 課題を付与した研究結果に即していた。一方で、健常人では関節運動制限の有無に関わらず、踵接地位置も時空間変数も左右対称性を確立していた。これらの結果を踏まえると、健常人では基本的にステップ長をはじめとした時空間変数が両側で対称な歩行パターンが定常であるため、split-belt での歩行適応時に対称化するよう収束すると考えられる。Finley ら [60]は、同様に split-belt 課題を用いて、片麻痺者の非対称歩行パターンが特定の個人にとって最適なパターンであり得ることを述べており、本研究結果もそれを支持している。つまり、健常人であっても有病者であっても、個人特有の歩行における定常性を持っており、その本質は左右対称であることではなく、両下肢が全身との関係において力学的な均衡性がとれていることと考えられた。本研究により得られた知見は、リハビリテーション介入にあたり、時空間変数の左右対称に近づけることを画一的な目標とすることに対し、再考の余地があることを指摘する。

9. 研究限界と今後の展望

Split-belt treadmill を用いた歩行学習の実験プロトコルでは、ベルト速度が非対称な条件で歩行パターンを学習させた後、再びベルト速度対称の条件に戻し、学習されたパターンの後効果をみる。本研究の主眼は、非対称歩行環境に適応する過程での歩行動態の変化における運動制御を分析することに置かれていた。今回、我々のいずれの研究もこの後効果をみる相を設けなかった。そのため、学習効果に関する直接的な言及を避けた。

本研究のデータには CoM の推定精度が関与している。CoM は、各体節に置いたマーカ位置情報と各身体慣性係数を用いて算出しており、かつ Vicon のマーカ位置情報の計測誤差は 5mm 程度とはいえ、実測した CoM と異なるという三次元動作解析の技術的限界が存在する。本研究の結果はあくまでも推定値である。また、矢状面上の 2 次元空間で動的安定性を推定したことで、次元削減により水平面情報が考慮されていない手法的限界がある。例示すれば、大きく外に開いて足を接地するといった情報は加味されないことになる。これらが重要な示唆を与える可能性は考えられ、今後 CoM と CoP の極座標系での推定などが望まれる。

研究Ⅱでは物理的な関節運動制限による影響を明らかにしたに過ぎず、脳卒中者は痙性による動的な運動制限、及び運動ニューロンレベルでの変化など、その他の神経的要因による影響については明らかにできていない。

本研究はヒトを対象とし、全ての対象者は大脳とそれ以下での連絡が保たれており、片麻痺者もその例外ではない。また、課題中に手すり使用により歩行適応に影響する可能性のある外部からの情報を得られた

可能性がある。逆に研究全体では外部情報を遮断するヘッドホンを使用しなかった。これらを含め、結果に大脳の随意的な制御による好影響または悪影響が含まれている可能性は完全に排除されていない。

研究Ⅲで脳卒中者のサブグループ化に伴い、対象者数が限られていた。被験者数が増えることで、今回分類した 2 グループ以外に区分される対象者がある可能性も存在する。そのため、臨床応用など、疾患者に対する示唆として一般化できる可能性は限定的である。さらに被験者を増やし、今回関連性の見えなかった重症度や罹患期間を加味することで、その追検証を行う必要がある。

最後に、片麻痺者における接地位置の対称性を再確立する能力について、直接的に言及できる実験系での検証を今後も目指している。

10. 結語

本研究は、CoM-CoP 角の対称性の再確立が split-belt treadmill が提供する非対称歩行環境への適応において必要な要素であることを示した。それは、接地した前脚と身体を力学的に有利な相対位置に配置することを、ベルト速度に依存せずに左右で同様に収束させる能力といえる。脳卒中片麻痺者においては、この能力を有するかが時空間変数の適応性を決定した。この結果は、split-belt treadmill を歩行運動学習のための介入ツールとして用いる目的の明確化と、適用となる対象の解明に貢献する可能性がある。

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1. Influence of arm joint limitation on interlimb coordination during split-belt treadmill walking

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Influence of Arm Joint Limitation on Interlimb Coordination during Split-belt Treadmill Walking

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Abstract During walking, arm and leg swings in healthy people are closely coupled temporally by interlimb coordination. A split-belt treadmill contains two belts that can be driven at different velocities, and is used to demonstrate the adaptability of human bipedal locomotion. Previous studies focusing on the use of split-belt treadmills in patients with neurological disorder demonstrated the existence of impaired temporal limb coordination in such patients, but the influences of neural and nonneural factors on interlimb coordination could not be examined separately. Further, the influence of limiting one joint of an arm on temporal coupling was unclear. The purpose of this study was to clarify the influences of limiting one arm joint and the corresponding compensation by the unlimited arm. Ten healthy young adults walked on a double-belt treadmill equipped with force sensors, during a tied-belt period (velocity of both belts = 0.9 m/s, 3 min) followed by a split-belt period (belt velocities = 0.9 m/s and 1.8 m/s, 6 min). The following experimental conditions were studied: slow side restrained, fast side restrained, and unrestrained (NR). A non-flexible bandage-type restraint limited the elbow extension to 20°. The correlations between the trajectories of arm and leg swings were analyzed using a Vicon motion capture system. The correlation coefficients between the restraint arm swing and slow- or fast-side leg swing were significantly lower in the restraint conditions compared to other condition in both tied-belt and split-belt periods. In particular, the anti-phase swing of the ipsilateral arm and leg and the in-phase swing of the contralateral arm and leg decreased. These results suggest that elbow limitation inhibits interlimb temporal coordination. The unrestrained arm swing increased in spatial amplitude, but maintained higher temporal coupling with the leg. Non-neural factors are expected to cause reduction of interlimb coordination in individuals having neurological disorders and joint limitations.

Keywords: walking, arm swing, restraint, split-belt treadmill, interlimb coordination.

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1. Introduction

Human bipedal locomotion involves spatial and temporal coordination in the motor output of arm and leg movements [1–4]. In particular, arm and leg swings are closely coupled temporally by interlimb coordination [5–8]. Previous studies have clarified that healthy people alternate between anti-phase and in-phase swings of the ipsilateral and contralateral arms and legs, respectively [9, 10]. Individuals with neurological disorders (such as post-stroke, Parkinson's disease, and multiple sclerosis) suffer from impaired limb coordination [11–15]. However, many patients possess spatially asymmetric bodies due to non-neural factors such as limited range of motion [16, 17]. Given these conditions, previous experiments performed on individuals with neurological disorders were not able to separate and clarify the neural and non-neural factors pertaining to temporal interlimb coordination.

Investigations of interlimb coordination using split-belt treadmills have recently been reported [3, 18, 19]. A split-belt treadmill is an experimental tool to demonstrate the adaptability of human bipedal locomotion based on a process in which a participant walks on a double-belt treadmill, the two belts of which can be driven at different speeds. Previous studies have clarified that temporal coupling between arm and leg swings is higher

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during split-belt treadmill walking [20]. Another study [21] using a split-belt treadmill has demonstrated that leg swings are markedly influenced by unilateral arm swings (induced by arm restriction or weight on the wrist). Therefore, split-belt treadmill walking is a useful experimental task that can help clarify the interlimb coordination in walking. However, the effect of limitation of a single joint in one arm on interlimb coordination has not yet been appropriately studied. Individuals with neurological disorders (such as post-stroke and Parkinson's disease) suffer passive joint limitation owing to chronic rigidity and spasticity. Before conducting experiments involving such patients, the influence of joint limitation, which is a non-neural factor, on interlimb coordination in healthy people without neurological disorders should be clarified.

To this end, this study aimed at examining the influence of partial restraint of an arm on interlimb coordination, to clarify the physical influence of gait control in patients suffering from joint limitation. Understanding the influence of arm joint limitation on interlimb coordination during walking is clinically significant for individuals having both neural and non-neural disorders. We hypothesized that restraint of a single joint in one upper limb influences lower coordination involving the contralateral arm and contralateral or ipsilateral legs.

2. Methods

Ten healthy young adults (males, age: 22 ± 1 years, weight: 60.5 ± 5.4 kg, and height: 1.75 ± 0.06 m) were recruited from the Saitama Prefectural University community. No participants had any history of neurological or musculoskeletal disorders. All participants provided written informed consent in accordance with the Declaration of Helsinki prior to the start of the proposed investigation. The study was approved by the ethics review committee in the Saitama Prefectural University.

2.1 Split-belt protocol

The experimental protocol is depicted in **Fig. 1**. All participants were instructed to walk on a double-belt treadmill, the two belts of which were equipped with force sensors (ITR5018-11, Bertec, US, 1000 Hz; **Fig. 2**). The two belts were controlled to operate at the same or different speeds by independent motors. The treadmill was operated under the following two conditions: symmetric condition (both belts moving at the same speed; "tied-belt") and asymmetric condition (two belts moving at different speeds; "split-belt"). The fast- and slow-moving sides of the treadmill during the asymmetric condition were termed "fast side" and "slow side," respectively. Both treadmill belts were set at the same speed of 0.9 m/s during symmetric operation (3 min). Subsequently, the speed of one belt was increased to 1.8 m/s with an acceleration of 0.5 m/s^2 during asymmetric operation (6 min).

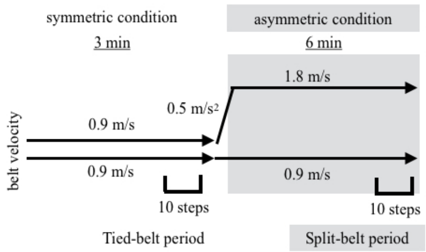

Fig. 1 Gait protocol and data collection periods.

Fig. 2 Experimental setup.

2.2 Joint limitation—elbow restraint condition

Experiments were performed in accordance with the abovementioned split-belt protocol under two conditions: with and without one-elbow restraint. The elbow restraint using a nonflexible elastic bandage (CB-25, NI-TREAT, Japan, **Fig. 2**) limited the range of motion to within 20–145°. This restraint did not limit flexion, and only extension was restrained to angular displacements within 0–20°. During elbow restraint, the order of assignment of the side with elbow restraint and the fast-side belt was randomized across participants. The following restraint conditions were examined: (1) non-restraint (NR), (2) slow-side elbow restraint (SR), and (3) fast-side elbow restraint (FR).

2.3 Data collection

Three-dimensional marker trajectory data were collected using a 17-camera Vicon motion-capture system (Oxford, UK; 100 Hz) with the Plug-in-Gait full-body AI model marker setting. The collected data were low-pass filtered at 5 Hz using a second-order zero-phase-lag Butterworth filter. Arm and leg swings were defined by their amplitude in the anteroposterior direction of the relative peripheral limb marker trajectories to the central limb marker. The trajectories of a wrist marker relative to a shoulder marker were used to determine the arm swing. Similarly, the trajectories of an ankle marker relative to the midpoint between the anterior and posterosuperior iliac spine markers were used to determine the leg swing. Gait events (heel contact and toe off) were defined based on the vertical component of the ground reaction force (Fz) followed by low-pass filtering at 5 Hz using a fourth-order zero-phase-lag Butterworth filter. Data were recorded during the first 30 s of symmetric operation (tied-belt period) and last 30 s of asymmetric operation (split-belt period). The data recorded over 20 strides were analyzed.

2.4 Data analysis and statistics

The mean elbow angle during each period was calculated, and a paired t-test was performed. Pearson correlation coefficients were calculated for right arm swing versus left arm swing, right leg swing versus left leg swing, right leg swing versus left arm swing, and left leg swing versus right arm swing, during each stride. The mean symmetrical ratio of arm amplitude was calculated during each period and under each condition. The symmetrical ratio was calculated using amplitudes of the fast- and slow-side arms. The mean ratio of arm amplitude during the split-belt period was normalized to that during the tied-belt period under each condition. A two-way repeated measures ANOVA was used to test for statistically significant differences in correlation coefficients between the restraint conditions (NR, SR, and FR) and periods (tied- and split-belt). A one-way repeated measures ANOVA was next used to compare the correlation coefficients and ratios pertaining to the restraint conditions (NR, SR, and FR) during each period (tied- and split-belt). When the ANOVA results yielded a significant effect, post hoc analyses were performed employing Bonferroni's test ($p < 0.05$). All analyses were performed using the MATLAB package (MathWorks Inc., USA).

3. Results

Table 1 lists the maximum extension angles of the elbow during walking. Significant differences ($p < 0.05$) were observed between the restrained and unrestrained sides under SR and FR conditions.

All correlation coefficients indicated that there were no interactions between the restraint conditions and periods [$F(2,18) = 0.34–2.29$, $p = 0.12–0.71$]. The differences in correlation coefficient between the slow- and the fast-side arm swings were not significant during both tied-belt and split-belt periods (**Fig. 3A**). Similarly, the differences in correlation coefficient between the slow-side and fast-side leg swings were not significant during both tied-belt and split-belt periods (**Fig. 3B**). The correlation coefficient increased significantly from tied-belt to split-belt periods in all conditions.

The correlation coefficient between the slow-side arm and fast-side leg swings during the tied-belt period was significantly smaller under the SR condition compared to FR during tied-belt as well as split-belt periods (**Fig. 3C**). The correlation coefficient increased significantly from tied-belt to split-belt period in all conditions. Likewise, the correlation coefficient between the slow-side arm and slow-side leg swings was significantly smaller under the SR condition compared to FR during

Table 1 Mean and standard deviation of the maximum extension angles (degree) of elbow joint during tied-belt and split-belt periods.

	Tied-belt			Split-belt		
	Restraint (NR = right)	Non-restraint (NR = left)	*p*-value	Restraint side (NR = right)	Non-restraint (NR = left)	*p*-value
NR	28.77 ±6.20	26.79 ±6.94	1	27.86 ±5.28	27.32 ±7.29	1
SR	44.28 ±18.84	27.30 ±7.30	<0.05	41.63 ±21.42	26.58 ±8.12	<0.05
FR	44.25 ±14.18	27.48 ±8.44	<0.05	45.03 ±19.29	26.50 ±7.99	<0.05

P values for SR and FR: differences between restraint and non-restraint sides. *P* values for FR: differences between left and right sides

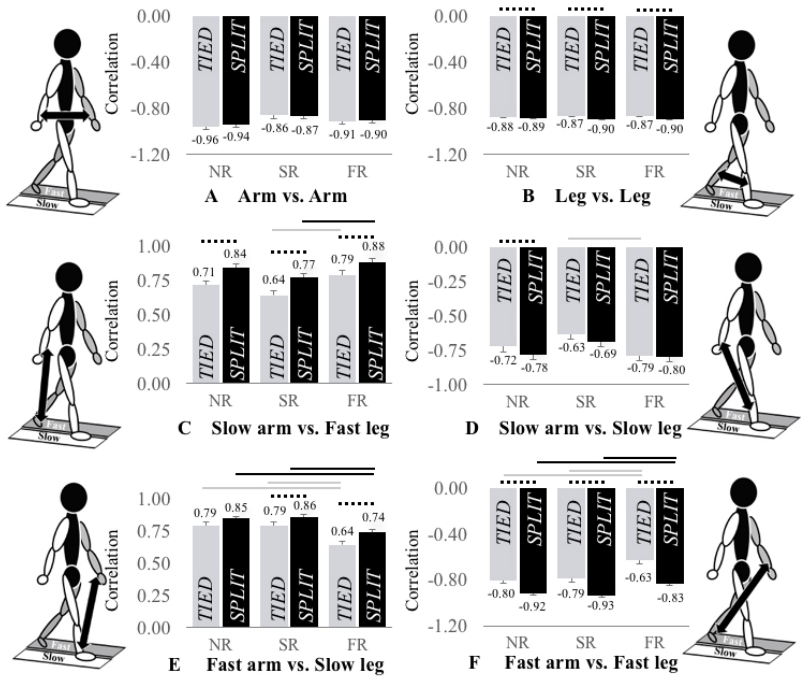

Fig. 3 Observed interlimb correlation coefficients (mean ± SD) during the tied-belt (gray) and split-belt (black) periods. Significant differences ($p > 0.05$) between conditions are denoted by horizontal bars. Horizontal dotted bars show significant differences ($p < 0.05$) between tied- and split-belt periods under different conditions. The correlation coefficients between both arms (A), both legs (B), slow side arm / fast side leg (C), slow side arm / slow side leg (D), fast side arm / slow side leg (E), fast side arm / fast side leg (F) are shown.

the tied-belt period (**Fig. 3D**). The corresponding difference during the split-belt period was, however, not significant (**Fig. 3D**). The correlation coefficient between the fast-side arm and slow-side leg swings was significantly smaller under the FR condition compared to those under the NR and SR conditions during the tied-belt and split-belt periods (**Fig. 3E**). The correlation coefficient between the fast-side arm and fast-side leg swings was significantly lower under the FR condition compared to those under the NR and SR conditions during the tied-belt and split-belt periods (**Fig. 3F**). The correlation coefficient increased significantly from tied-belt period to split-belt period in all conditions.

Figure 4 depicts the mean symmetrical ratios of arm amplitudes under each of the three conditions—NR, SR, and FR. During the tied-belt period, no significant difference in amplitude of the unrestrained arm was observed between the NR (1.18), SR (2.32), and FR (2.29) condi-

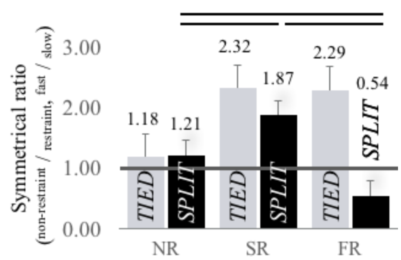

Fig. 4 Mean symmetrical ratio of arm swing amplitudes during tied-belt (gray; unrestrained/restrained) and split-belt (black; fast/slow) periods under each condition. Significant differences ($p > 0.05$) between conditions are denoted by horizontal bars.

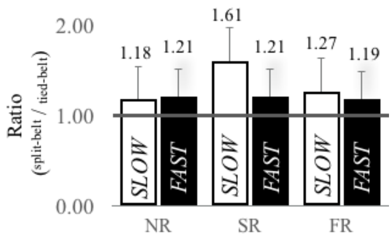

Fig. 5 Mean ratio of arm swing amplitude during split-belt period normalized to that during tied-belt under each condition. White columns denote the slow side arm swing while black columns correspond to the fast side. No significant difference was observed between conditions.

tions. During the split-belt period, the amplitude of the fast-side restrained arm was small (0.54) under the FR condition and was large (1.87) under the SR condition. During this period, the symmetrical ratios were significantly different between all three conditions.

Figure 5 depicts the mean ratios of arm amplitudes in the split-belt period normalized to the corresponding values in the tied-belt period. No significant difference in ratio was observed between all three conditions for the slow side (NR = 1.18, SR = 1.61, FR = 1.27) and the fast side (NR = 1.21, SR = 1.21, FR = 1.19).

4. Discussion

4.1 Inhibition of interlimb temporal coordination by arm restraint

As some correlation coefficients showed significant increases from the tied-belt to the split-belt period, the walking adaptation phenomenon occurred partially under restrained conditions. The other results obtained in this study demonstrated that the correlation coefficients between the restrained arm swing and slow- or fast-side leg swing were significantly smaller under restrained conditions compared to other condition during both tied-belt and split-belt periods (SR in **Figs. 3C and 3D**; FR in **Figs. 3E and 3F**). In particular, anti-phase swing of the ipsilateral arm and leg and in-phase swing of the contralateral arm and leg were reduced during both periods. These results indicate that elbow limitation tends to inhibit interlimb temporal coordination. Locomotion is based on the operation of central pattern generators, which are defined as neural networks capable of producing coordinated patterns of rhythmic limb activity [22]. Interlimb coordination usually manifests via ipsilateral neural connections between the upper and lower limbs [23]. Based on the contribution of the central nervous system, MacLellan [20] reported that the temporal coupling between arm and leg swings is stronger during split-belt treadmill walking. For example, post-stroke patients and those with Parkinson's disease experience contractures in arm joints owing to rigidity or spasticity of biceps brachii [24], similar to the restraint conditions examined in the present study. This is because such contractures are typically limited to manifestation along the extension direction. Individuals having such neurological disorders likely possess physical restraint factors similar to the model examined in this study, in addition to neural factors. Based on this result, such patients may also be affected by inhibition of interlimb coordination owing to physical constraints.

4.2 Emphasis on unrestrained side arm swing

The arm amplitudes on the slow and fast sides during the split-belt period differed only slightly compared to those in the tied-belt period under each condition (**Fig. 5**). However, during the tied-belt period, the symmetrical ratio (unrestrained / restrained side) was much greater than 1.00 in both SR and FR conditions (**Fig. 4**). In addition, during the split-belt period, the symmetrical ratio (fast / slow side) was greater than 1.00 in SR and smaller than 1.00 in FR (**Fig. 4**). This finding also indicates that the arm swing amplitude on the unrestrained side became greater than that on the restrained side. Therefore, in the present study, the arm swing amplitude on the unrestrained side was observed to increase regardless of the side being slow or fast. At a speed ratio of 2:1 during split-belt walking, MacLellan et al. [20] demonstrated an increase in arm swing amplitude exclusively on the slow side, and not on the fast side. They explained that upper-limb amplitudes correspond to those of the fast-moving lower limb, thereby indicating an adaptation to maintain symmetric upper limb movements governed by the central nervous system, as proposed in the lower limbs [18, 20]. Under the restraint conditions examined in this study, the influence of restraint on upper limbs led to compensatory increase in swing of the contralateral arm. Moreover, the observed compensatory effect was stronger compared to that of the difference in belt speed.

During split-belt walking with elbow restraint, the fast-side arm could swing emphatically to compensate for the restraint. Bondi et al. [21] performed a study on arm swing using the split-belt treadmill and examined the conditions of excessive arm swinging and weight attachment on the arm. They observed an increase in leg swing time under the condition of excessive arm swinging, but a tendency of decrease in the case of weight attachment to the arm. An interpretation of these results is that the observed increase or decrease in rhythmic arm swing amplitude is transferred to the lower limbs mediated by the overall supra spinal involvement, thereby affecting symmetric leg swings. The findings reported by Bondi et al. [21] support the results obtained in our study concerning the influence of arm joint limitation on interlimb coordination.

4.3 Compensation of unrestrained side arm swing

As the spatial amplitude of arm swing on the unrestrained side became larger, temporal coupling with the leg increased (FR values in **Figs. 3C and 3D**; SR values in **Figs. 3E and 3F**). Previous studies using split-belt walking demonstrated that trunk movement also becomes tightly coupled to the leg [20, 25], suggesting that the leg directly drives trunk rotation through biomechanical linkages and the arm swing is used to balance the trunk torques. Therefore, in individuals experiencing passive limitation in one arm, the contralateral arm may continue to demonstrate temporal coupling with both legs. Therefore, in individuals experiencing joint limitations and movement disorders in the arm and leg on the same side, such as in post-stroke patients, the above result suggests that such individuals may swing the nonparetic arm in a compensatory manner to facilitate easy swing of the paretic leg.

5. Limitation

The present study had a methodological limitation, in that only young male adults were recruited for this study. It is, therefore, unclear whether findings of this study could be generalized to other populations. Future investigations should include post-stroke patients to verify the clinical application of the proposed concept.

6. Conclusion

The present study demonstrates that for arm swing during split-belt walking, joint limitation of one elbow inhibits the temporal interlimb coordination between the ipsilateral arm and both legs. As the spatial amplitude of the contralateral arm swing becomes greater, temporal coupling with the leg increases. Based on this result, individuals with neurological disorders and passive joint limitation may experience inhibition of interlimb coordination owing to physical constraints.

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2. Verification of the adaptive parameters of the relative positions of the leading leg and the whole body at foot contact during split-belt treadmill walking

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Verification of the adaptive parameters of the relative positions of the leading leg and the whole body at foot contact during split-belt treadmill walking

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Abstract: The main purpose of this study was to demonstrate that adaptive walking re-establishes dynamic stability at the time of foot contact in split-belt treadmill walking. First, we examined whether the relative positions of the center of mass (COM) and the center of pressure (COP) were re-established at foot contact during walking adaptation. We found that the symmetry of the COM-COP angles was re-established more completely than any other parameters (foot and pelvis markers) on both the fast and slow belt sides during the adaptation phase. Moreover, we examined whether the dynamic relationship between the COM and the COP was the same during tied- and split-belt walking. Our results showed that the equations related to the trailing leg relative to the COM re-converged around 0 at foot contact on both the fast and slow sides during the adaptation phase. We also showed that the COM-COP angle at foot contact is an adaptive parameter and is consistent between the fast and slow belt sides. At foot contact, the relative positions between the leading leg and the whole body can also be seen to be adjusted during trial-and-error adaptation to split-belt perturbation in which the belts move at different speeds. Therefore, we suggest that the relative positions of the COM and the COP at foot contact is the spatial reference for the gait cycle contribution to dynamic stabilization during adaptive walking.

1. INTRODUCTION

Able-bodied people can alter their walk to adapt to various environments through the generation of usable motor commands in a neural system. Movement is generated by motor commands coordinating between the legs (interlimb coordination) and between the joints of each leg (intralimb coordination). This coordination helps us control the movements of numerous joints. To demonstrate the underlying mechanism of interlimb and intralimb coordination, a recent study has investigated the adaptability of human bipedal locomotion through spatiotemporal changes under tied-belt (belts move at the same speed) and split-belt (belts move at different speeds) conditions [1]. During split-belt walking, interlimb (i.e., step length, double support time) and intralimb (i.e., stride length, stance time) spatiotemporal parameters show different adjustments. Interlimb parameters gradually adapt and re-establish symmetry during the split-belt period. Conversely, intralimb parameter adjustments are merely reactive and maintain asymmetry during split-belt perturbation [2].

A previous study using a split-belt treadmill showed that the foot contact timings are modified to be phasic center of gait cycle of both the slow and fast belt sides in adaptive locomotion [3]. Physiological and robotics studies have shown that locomotor rhythm and its phases are modulated by the production of phase shifts and rhythm resetting based on sensory afferents and perturbations (phase resetting) [4,5]. In order to influence periodic walking patterns, the foot contact timings of both the fast and slow sides must be dynamically stable. Quantifying dynamic stability during walking requires an understanding of how the center of mass (COM) and center of pressure (COP) movements are generated and controlled [6]. A previous study of falling suggest that concurrent interpretation of the COM and the COP of the leading leg can give a more complete assessment than examining them separately, because we often slip on the leading leg at foot contact and require dynamic balance [7]. However, few split-belt studies have calculated the COM by capturing the whole human body. Moreover, few studies considered dynamic stability during adaptive walking by examining the relationship between COM and COP.

The main purpose of this study was to demonstrate that adaptive walking re-establishes dynamic stability at the time of foot contact of the fast and slow belt sides during split-belt treadmill walking. First, we verified that the relative positions of the COM and the COP were re-established on the fast and slow sides via walking adaptation. We investigated the inclination angle between the COM and the COP of the leading leg. In addition, we researched other parameters to clarify the relationship between the leading leg and the whole body. Second, we examined whether the dynamic relationship between the COM and the COP was the same during tied- and split-belt walking. We investigated the effect of the trailing leg at the peak braking force using the pendulum model with COM and COP.

† **Keisuke Hirata is the presenter of this paper.**

2. METHODS

2.1 Subjects

Fifteen young adult male subjects participated in this experiment (age: M = 22.2 years, SD = 1.1). All subjects were recruited from among the students at Saitama Prefectural University. Individuals with a history of myocardial infarction, stroke, fracture, or symptomatic arthritis of the lower extremities were excluded from this study. All participants provided their informed consent prior to participating in the experiments. This study was conducted in accordance with the Declaration of Helsinki. The study was approved by the ethics review committee in the Saitama Prefectural University.

2.1 Experimental protocol

The participants walked on the treadmill following a protocol used in a similar experiment [8] (Fig. 1): (1) tied-belt at 0.9 m/s for 3 min followed by (2) split-belt with the slow belt at 0.9 m/s and the fast belt at 1.8 m/s (2:1 ratio of fast to slow) for 6 min. The fast and slow belt sides were randomly assigned to either their right or left leg.

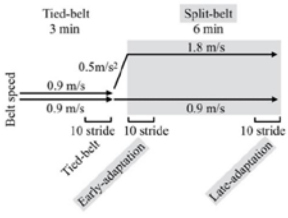

Fig.1 Split-belt protocol.

2.3 Data collection

A camera motion capture system (Vicon Motion Systems, Oxford, UK) was used to capture the three-dimensional movement and kinematic data (Fig. 2). We used 22 cameras and 39 reflective markers from the Plug-in-Gait full-body AI model to calculate the body's center of mass. Data were sampled at a rate of 100 Hz. The forces exerted on the instrumented double-belt treadmill (Bertec Corp., USA) were also measured (1000 Hz) and resampled to match the kinematic data. Recordings of the kinematic and kinetic signals were conducted for 30 seconds at the end of the tied-belt condition (150–180s, tied-belt phase), at the beginning of the split-belt condition (180–210s, early-adaptation phase), and at the end of the split-belt condition (510–540s, late-adaptation phase).

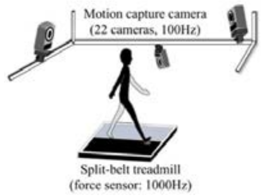

Fig.2 Motion capture system and split-belt treadmill.

2.4 Spatiotemporal parameters

We calculated the following spatiotemporal gait parameters. (1) Step length: the anteroposterior distance between the ankle markers of each leg at foot contact. (2) Percent stance time: the duration from the point of foot contact to toe off, expressed as a percentage of the stride time. (3) Percent double support time: the duration from the foot contact to toe off, expressed as a percentage of the stride time.

2.5 Parameters of the relative positions of the leg and whole body

We chose the ankle marker (ANK) and the COP as the parameters of the leading leg at foot contact. We chose the midpoint of the ipsilateral pelvis (iPEL), the center of the pelvis (PEL), and COM as the parameters of the whole body at foot contact. The iPEL was calculated as the midpoint between the anterior superior iliac spine (ASIS) and the posterior superior iliac spine (PSIS) on the side of foot contact. The PEL was calculated as the center of the ASISs and the PSISs on both sides. Moreover, we calculated the angle between a vertical line (against earth vertical) and the vector from the body parameter to the leg parameter on the sagittal plane. The angles calculated by combining the parameters were as follows: (1) iPEL-ANK angle, (2) iPEL-COP angle, (3) PEL-ANK angle, (4) PEL-COP angle, (5) COM-ANK angle, and (6) COM-COP angle.

2.6 Equations related to the trailing leg

The peak braking force of the leading leg, which is calculated from the vertical and anteroposterior components of the ground reaction force, usually occurs shortly after foot contact in the leading foot. This is an important factor related to slipping during walking known as the required coefficient of friction (RCOF). Previous studies clarified that the tangent of the COM-COP angle strongly correlates with the RCOF during walking [9] by the following calculations.

$$Mx = Fz1(y_{COP1} - y_{COM}) + Fz2(y_{COP2} - y_{COM}) - Fy1 * Z_{COM} + Fy2 * Z_{COM} \quad (1)$$

$$\frac{Fy1}{Fz1} = tan\theta + RT \quad (2)$$

$$tan\theta = \frac{y_{COP1} - y_{COM}}{Z_{COM}} \quad (3)$$

$$RT = \frac{Fz2}{Fz1} * \frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} + \frac{Mx}{Fz1 * Z_{COM}} \quad (4)$$

Yamaguchi et al. [9] estimated the moment around COM [*Mx*; Eq. (1)] using bipedal inverted pendulum modeling with COM and COP in the sagittal plane (Fig. 3). From Eq. (1), they derived the braking force for the leading leg [*Fy1/Fz1*; Eq. (2)]. Eq. (2) divides the tangent of the COM-COP angle of the leading leg [*tanθ*; Eq. (3)] and the equations related to the trailing leg, referred to as the residual term [*RT*; Eq. (4)]. Their results clarified that the peak braking force of the leading leg is approximately equal to the tangent of the COM-COP angle at the point of peak braking force on the leading leg (RCOF). Therefore, the equations related to the trailing leg were found to be very small at the moment of RCOF.

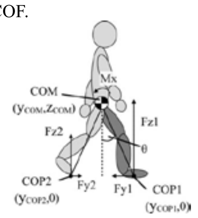

Fig.3 Bipedal inverted pendulum model.

To achieve our second objective, we verified the change in RT, symmetrical index of the RT, RCOF timing, and percent double support time during adaptation to split-belt perturbation.

2.7 Statistics

We calculated symmetrical ratios for all parameters as follows, based on a previous study [10].

$$\frac{fast\ side\ -\ slow\ side}{fast\ side\ +\ slow\ side} \times 100(\%) \qquad (5)$$

We compared each parameter during the last 10 strides (tied-belt phase and late-adaptation phase) and the first 10 strides (early-adaptation phase). To compare the results of each parameter, we used a repeated-measures analysis of variance (ANOVA) with a within subject factor of testing phase (tied-belt, early-adaptation, late-adaptation). The significance level was set at p = 0.05 which was adjusted using a Bonferroni correction for post hoc multiple comparisons. For the late-adaptation phase, a repeated-measures ANOVA was used to test the parameter comparisons (iPEL-ANK, iPEL-COP, PEL-ANK, PEL-COP, COM-ANK, COM-COP), with the significance level at p = 0.05, which was adjusted using a Fisher LSD correction for post hoc multiple comparisons. Statistical comparisons were completed using MATLAB and Statistica (StatSoft, Tulsa, OK) software.

3. RESULTS

3.1 Spatiotemporal parameters

Fig. 4 shows the time series changes in the symmetrical index of double support time, step length, and stance time for the comparison between conditions. Stance time was significantly different in both the early-adaptation (M = –10.66, SD = 2.95; p < 0.001) and late-adaptation (M = –8.44, SD = 10.00; p < 0.001) phases compared to that in the tied-belt phase (M = 0.41, SD = 1.15). Stance time remained asymmetrical during the split-belt condition.

Step length was significantly different (p < 0.001) between the early-adaptation phase (M = –21.57, SD = 8.11) and the tied-belt phase (M = 1.54, SD = 7.37). There was no significant difference between the late-adaptation phase (M = 0.64, SD = 5.97) and the tied-belt phase (p = 1.00). Symmetry in step length was re-established during the split-belt condition.

Similarly, for double support time, there was a significant difference between the tied-belt phase (M = –0.10, SD = 1.23) and the early-adaptation phase (M = 16.70, SD = 2.82; p < 0.001) but not the late-adaptation phase (M = 1.03, SD = 1.80; p = 1.00). Symmetry in double support time was also re-established during the split-belt condition.

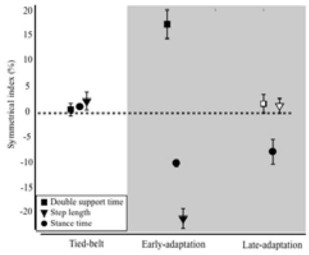

Fig.4 Comparisons of each spatiotemporal parameter at different time points. Filled circles indicate statistically significant differences compared the tied-belt phase. Error bars indicate mean standard error.

3.2 Parameters of the relative positions of the leg and whole body

Fig. 5 shows the time series changes in the symmetrical index of the iPEL-ANK, iPEL-COP, PEL-ANK, PEL-COP, COM-ANK, and COM-COP angles for the comparison between conditions.

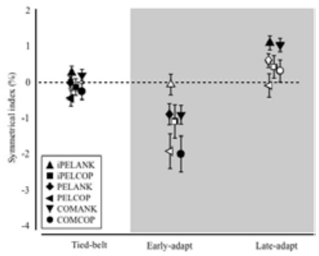

Fig.5 Comparisons of each parameter at different time points. Filled points indicate statistically significant differences compared to the tied-belt phase. Error bars indicate mean standard error.

There were no significant differences in the iPEL-COP and PEL-COP angles between the tied-belt phase (iPEL-COP: M = –0.16, SD = 0.93, PEL-COP: M = –0.51, SD = 0.96) and the early-adaptation (iPEL-COP: M = –1.10, SD = 1.85, p = 0.18; PEL-COP: M = –1.72, SD = 1.88, p = 0.06) and late-adaptation (iPEL-COP: M = 0.39, SD = 1.29, p = 0.77; PEL-COP: M = –0.10, SD = 1.47, p = 1.00) phases.

For the iPEL-ANK angle, the difference was not significant between the tied-belt phase (M = 0.23, SD = 0.72) and the early-adaptation phase (M = –0.05, SD = 1.20; p = 1.00), but was significant between the late-adaptation phase (M = 1.07, SD = 0.81) and the tied-belt phase (p <0.05). The iPEL-ANK angle increased in asymmetry during the split-belt condition.

The COM-ANK angle was significantly different in both the early-adaptation (M = –0.86, SD = 1.06; p < 0.01) and late-adaptation (M = 1.03, SD = 0.75; p < 0.01) phases compared to that in the tied-belt phase (M = 0.16, SD = 0.60). The asymmetry in the COM-ANK angle varied between the beginning and the end of the split-belt condition.

For the PEL-ANK and COM-COP angles, the difference was significant between the tied-belt phase (PEL-ANK: M = –0.03, SD = 0.63; COM-COP: M = –0.32, SD = 0.96) and the early-adaptation phase (PEL-ANK: M = –0.85, SD = 0.60, p < 0.01; COM-COP: M = –1.80, SD = 1.96, p < 0.001), but not between the tied-belt phase and the late-adaption phase (PEL-ANK: M = 0.60, SD = 0.78, p = 0.15; COM-COP: M = 0.29, SD = 1.34, p = 0.67). Symmetry in the PEL-ANK and COM-COP angles was re-established during the split-belt condition.

Fig. 6 shows a time series of changes of the COM-COP and iPEL-ANK angles for each side of foot contact. It is a single-subject example of step symmetry plotted stride-by-stride. Among the parameters of the relative positions of the leg and the whole body, the COM-COP and iPEL-ANK angles showed the clearest example of adaptation in both legs, and this pattern is symbolic behavior. In the beginning of the split-belt condition, both the COM-COP and iPEL-ANK angles indicated asymmetry that changed with increasing stride number. At the end of the split-belt condition, the COM-COP angles had returned to near tied-belt condition values and symmetry had been re-established (Fig. 6A). On the other hand, the iPEL-ANK angles had not returned to the tied-belt condition values, and asymmetry had expanded during the split-belt condition (Fig. 6B).

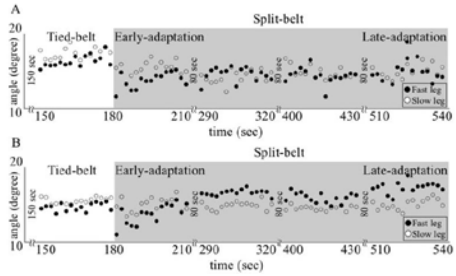

Fig.6 Time series of changes in COM-COP (A) and iPEL-ANK (B) angles of a single subject plotted stride-by-stride. Filled circles denote the angle of the fast leg at the fast side foot contact. Open circles denote the angle of the slow leg at the slow side foot contact.

Fig. 7 shows the comparisons of each parameter in the late-adaptation phase. The PEL-COP angle was significantly different to the COM-ANK and iPEL-ANK angles (p < 0.001) and the COM-COP angle was significantly different from the iPEL-ANK angle (p < 0.01).

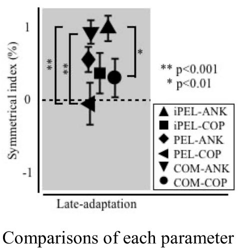

Fig.7 Comparisons of each parameter in the late-adaptation phase. * indicates p-value < 0.01, ** indicates p-value < 0.001. Error bars indicate mean standard error.

3.3 Equations related to the trailing leg

Table 1 shows the symmetrical index of the RT for each leg and condition. There was no significant change in the RT at the point of fast side foot contact in the early- and late-adaptation phases (p = 1.00) compared to that in the tied-belt condition. For the RT at the point of slow side foot contact, there was a significant increase between the tied-belt condition and the early-adaptation phase (p < 0.001), but not between the tied-belt condition and the late-adaptation phase (p = 1.00). Similarly, the symmetrical index was significantly different in the early-adaptation phase (p < 0.001) but not in the late-adaptation phase (p = 0.07) compared to that in the tied-belt condition.

Table 1 also shows the timing of RCOF and the duration of double support time as a percentage of normalized stance time. There were no significant differences in the timings of RCOF between any phases for either leg (*p* = 1.00; approx. 12% per stance time). Compared to the tied-belt phase, the percent double support time was significantly shorter for the fast leg ($p < 0.001$) and longer for the slow leg ($p < 0.01$) during the early-adaptation phase and significantly shorter for the fast leg in the late-adaptation phases.

Table 1 Descriptive statistics (Mean ± SD) for the equations related to the trailing leg: a residual term (RT), symmetrical index of the RT, the required coefficient of friction (RCOF) timing, and percent double support time. Except for the symmetrical index, the results are shown for each leg.

	Tied-belt		Split-belt			
			Early-adaptation		Late-adaptation	
	Fast	Slow	Fast	Slow	Fast	Slow
RT	0.09 ± 0.03	0.10 ± 0.04	0.08 ± 0.03	0.14 ± 0.04**	0.08 ± 0.03	0.10 ± 0.03
Symmetrical index of RT (%)	1.57 ± 12.72		30.42 ± 12.08**		13.74 ± 19.15	
RCOF timing (%)	12.15 ± 2.39	12.78 ± 2.17	12.42 ± 1.41	12.56 ± 2.60	11.68 ± 1.78	12.79 ± 2.97
Percent double support time (%)	18.59 ± 1.23	18.52 ± 2.36	12.72 ± 1.98**	21.61 ± 2.62*	15.53 ± 3.62*	18.49 ± 2.50

Asterisk indicates statistically significant differences compared to the tied-belt phase.
* indicates *p* value < 0.01, ** indicates *p* value < 0.001.

4. DISCUSSION

The main purpose of this study was to demonstrate that adaptive walking re-establishes the dynamic stability at the point of foot contact in split-belt treadmill walking. First, we examined whether the relative positions of the COM and the COP were re-established at foot contact on the fast and slow sides of the belts via walking adaptation. We showed that symmetry in the COM-COP angles was re-established on both the fast and slow sides during the phase of adaptation to the split-belt condition. Second, we examined whether the dynamic relationship of the COM and the COP was the same during tied- and split-belt walking. We found that equations related to the trailing leg relative to the COM re-converged around 0 at foot contact on both the fast and slow sides during the adaptation phase.

Table 1 also shows the timing of RCOF and the duration of double support time as a percentage of normalized stance time. There were no significant differences in the timings of RCOF between any phases for either leg (*p* = 1.00; approx. 12% per stance time). Compared to the tied-belt phase, the percent double support time was significantly shorter for the fast leg ($p < 0.001$) and longer for the slow leg ($p < 0.01$) during the early-adaptation phase and significantly shorter for the fast leg in the late-adaptation phases.**.1 Re-establishing symmetry in the COM-COP angle**

First, the spatiotemporal gait parameters showed that walking adaptation to the split-belt condition occurred in this experiment [2]. Symmetry of the interlimb parameters (step length and double support time) changed over time between the early- and late-adaptation phases. The intralimb parameter (stance time) became asymmetric just after the shift to the split-belt condition and remained asymmetrical in the late-adaptation phase.

In the parameters concerning the relative positions of the leading leg and the whole body, clear symmetry of the COM-COP angle was re-established in the late-adaptation phase. Previous studies have demonstrated that the iPEL-ANK angle is similar to the limb angle (the angle between a vertical line and the vector from the greater trochanter to the ankle marker) [11, 12]. Malone et. al. demonstrated that the limb angle at foot contact was asymmetrical during late-adaptation; our results for the iPEL-ANK concur with this finding [1]. However, although the COP-COM angle is similar to the iPEL-ANK and limb angles, the symmetry of this angle was re-established during the late-adaptation phase in our study. The relative positions of the COM and the COP at foot contact was consistent between the fast and slow sides and was not dependent on the belts' speeds. In terms of dynamic stability during walking, the interaction of foot placement, COM, and COP at foot contact is crucial [7]. The tangent of the COM-COP angle is similar in young adults and elderly people at the point of RCOF [13], because of the following factors: step length, COM velocity, and COM position. It suggests that able-bodied people have a consistent relative position between the leading leg and the whole body to maintain dynamic stability when walking. Our examination of the re-establishment of COM-COP angle symmetry suggests that foot contact controls dynamic stabilization during split-belt walking.

Table 1 also shows the timing of RCOF and the duration of double support time as a percentage of normalized stance time. There were no significant differences in the timings of RCOF between any phases for either leg (*p* = 1.00; approx. 12% per stance time). Compared to the tied-belt phase, the percent double support time was significantly shorter for the fast leg ($p < 0.001$) and longer for the slow leg ($p < 0.01$) during the early-adaptation phase and significantly shorter for the fast leg in the

late-adaptation phases..2 Re-convergence of the equations related to the trailing leg

In our results, the equations related to the trailing leg relative to the COM re-converged around 0 at the time of RCOF. This result indicated that the tangent of the COM-COP angle was close to the value obtained by the dividing braking force by the vertical force (RCOF) of the leading leg at weight acceptance during the adaptation phase. The difference between the tangent of the COM-COP angle and the RCOF is increased when dynamic stability is decreased, as in slow walking [9]. This occurs when the moment from the leading and trailing leg to the COM (Mx) is not 0 [9]. Therefore, dynamic stability at weight acceptance is re-established between the early- and late-adaptation phases. Moreover, the peak braking force at weight acceptance is the interlimb parameter, because the effect of the trailing leg is canceled. Our argument is supported by a previous study that found that the magnitude of braking force was the adaptive parameter [14].

We found that the RCOF timings and percent double support time per gait cycle were always 11-12% and 15-18%, respectively. This is the time from foot contact to weight acceptance (0-10%) when the body's center of mass is propelled forward toward the forefoot [15]. Reisman argues that adapting the angle of the leading leg at weight transfer is critical for controlling the relative positions of the COM and the leading leg [2]. Therefore, in finding that the relative positions of the COM and the COP were consistent at foot contact on both the fast and slow sides and did not depend on the belts' speeds., our study supports this argument.

This study had some limitations. First, we recruited only young male adults, making it unclear whether our findings can be generalized to other age and sex classes. Second, showing the after-effects on the parameters during the post-adaptation phase would have allowed us to investigate motor learning. Our protocol did not include a second tied-belt condition phase after the split-belt condition (the post-adaptation phase). Therefore, we were only able to discuss adaptive behavior during the split-belt condition.

5. CONCLUSION

This study showed that the COM-COP angle at foot contact is an adaptive parameter and is consistent between the fast and slow belt sides. This phenomenon may occur in concurrently with re-establishing dynamic stability during split-belt treadmill walking. Previous studies mention that the interlimb parameter that represents adaptation is controlled predictively [2,11]. At foot contact, the relative position between the leading leg and the whole body can also be seen to adjust itself during trial-and-error adaptation to split-belt perturbation, that is, a different speed for each belt. We suggest that the relative positions of the COM and the COP at foot contact is the spatial reference in the gait cycle contribution to dynamic stabilization during adaptive walking.

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3. Adaptive changes in foot placement for split-belt treadmill walking in individuals with stroke

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Adaptive changes in foot placement for split-belt treadmill walking in post-stroke patients

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None of the authors have any conflicts of interest associated with this study.

ABSTRACT

Background: Adaptation to split-belt treadmill walking differs between individual stroke survivors. Many discussions address only spatiotemporal parameters that are related to movement, and the changes in interlimb spatiotemporal parameters as a consequence of adaptation are poorly understood.

Objectives: To investigate symmetry of the center of pressure (CoP) position relative to the center of mass (CoM), and ascertain whether differences due to adaptation can be identified by studying interlimb spatiotemporal parameters during split-belt walking.

Methods: Twenty-two chronic post-stroke patients and nine elderly and seventeen young adult controls walked in tied- then split-belt (2:1 ratio of fast:slow) conditions. Spatiotemporal parameters were compared within groups to assess symmetry of the CoM-CoP angle at foot contact.

Results: Asymmetry of the CoM-CoP angle was associated with asymmetry of spatiotemporal parameters. Re-establishment of symmetry of CoM-CoP angle was reflected in re-established symmetry of spatiotemporal parameters in post-stroke and control participants.

Conclusions: Post-stroke patients who re-establish symmetry of the COM-COP angle are able to adapt their walking for split-belt perturbation. This suggests that predictively

symmetric foot placements on the fast and slow sides are necessary for adaptation in walking. Symmetrical foot placement is achieved by interlimb coordination and may contribute to dynamic stability.

2

1. Introduction

Post-stroke patients typically walk asymmetrically and assessment of spatiotemporal and kinetic variables in such individuals will reveal differences between the paretic and non-paretic limbs. Examples of asymmetrical variables include stance and swing times [Wall et al, 1986], double-support time [Olney et al, 1994], joint power [Knutsson et al, 1979], joint excursion [Olney et al, 1991], and step length [Allen et al, 2011]. Asymmetry of post-stroke patients is thought to be caused by various obstacles to locomotion [Patterson et al, 2008; Patterson et al, 2010], and split-belt treadmill training has recently emerged to induce the learning of a new walking pattern with reduced asymmetry of step length and double-support time [Reisman et al, 2007; Reisman et al, 2013; Betschart et al, 2017; Miéville et al, 2018]. This is achieved by improvement in interlimb coordination due to patients' adaptability [Tyrell, 2015; Wutzke, 2015].

Split-belt treadmills use separate double belts to drive the legs at different speeds, and the adaptation protocol is used to demonstrate the adaptability of human bipedal locomotion by time-series changes in the symmetry or asymmetry of gait spatiotemporal parameters [Dietz et al, 1994; Choi et al, 2007]. During split-belt walking, interlimb (i.e., step length, double-support time) and intralimb (i.e., stride length, stance time) spatiotemporal parameters show different changes. Interlimb parameters gradually adapt and re-establish symmetry, whereas intralimb parameters adjust merely reactively, and asymmetry is maintained [Reisman et al, 2005].

However, not all post-stroke patients completely re-establish symmetry during split-belt walking. Previous studies have reported that the magnitude [Savin, 2013], time [Tyrell, 2015], and reaction manner [Reisman et al, 2007; Malone et al, 2014] of interlimb spatiotemporal parameters are different depending on the individuals patient and belt conditions (i.e., paretic side slow or fast). Because many studies have focused on spatiotemporal parameters that occur as a result of movement, the causes of differences in adaptation of interlimb spatiotemporal parameters following split-belt walking are unclear. Kinetic analysis of gait adaptation in split-belt conditions may provide clarification of these causes. For adaptation during split-belt treadmill walking, adequate control of the position of the center of mass (CoM) contributes to stability during locomotion [Morton, 2006; Mawase et al, 2013; Jansen et al, 2013]. Moreover, studies of split-belt treadmill training in post-stroke patients have shown that foot placement affects dynamic stability [Hak, 2013; Reisman, 2013; Miéville, 2018]. We hypothesized that differences in the adaptation of interlimb parameters are related to dynamic stability in post-stroke patients.

Dynamic stability during walking relies on the interactions that affect foot placement, and the positions of the CoM and center of pressure (CoP) at the moment of foot contact are crucial [Hsue et al, 2009]. In our preliminary study of the inverted pendulum model of CoM and CoP, healthy subjects re-established symmetry of leading-foot placement relative to the whole body at foot contact during split-belt walking [Hirata et al, 2019a in submission]. The present study aimed to verify that symmetry of the CoP position relative to CoM can indicate differences in adaptation of interlimb spatiotemporal parameters in post-stroke patients during split-belt walking. This verification could add kinetic explanations of the differences that are observed in the dynamic stability adaptability of post-stroke patients.

2. Methods

2.1. Participants

Twenty-two individuals who had sustained a single stroke more than 6 months prior to the study (mean age: 67.0 years, standard deviation [SD]: 9.3; 17 females, Table 1), nine age- and gender-matched healthy elderly control subjects (0.4 matched controls per case), and seventeen healthy young adult control subjects (mean age: 22.1 years, SD: 1.2; 17 females) were recruited to participate in the study. Exclusion criteria were: other neurological conditions, orthopedic conditions affecting the legs or back, uncontrolled hypertension, pacemaker or automatic defibrillator fitted, active cancer, evidence of damage to the cerebellum by radiological and/or physical examination, or inability to complete the task. Subjects who customarily wear an ankle-foot orthosis were allowed to wear it during testing. The experiments were explained in detail to participants, and written informed consent obtained. This study and its protocols were approved by the Saitama prefectural university ethics review committee (No. 29501).

2.2. Clinical evaluation

Prior to the experiments, participants walked on the treadmill to determine their comfortable and maximum walking speeds. Lower-limb motor recovery was evaluated with the lower extremity portion of the Fugl-Meyer Assessment (LE FMA) [Fugl-Meyer et al, 1975]. The Timed Up and Go test [Ng et al, 2005] and Functional Independence Measure (FIM) were also carried out.

2.3. Experimental protocol

The split-belt treadmill protocol followed a published protocol [Hirata et al, in press] (Figure 1A). Post-stroke and healthy elderly participants walked on the treadmill under the following conditions: (1) "tied-belt" at half-maximum walking speed for 3 min, then (2) "split-belt" with the slow belt at half-maximum and fast belt at maximum walking speed (2:1 ratio of fast:slow) for 6 min. Healthy young participants walked under the following conditions: (1) tied-belt at 0.9 m/s for 3 min, then (2) split-belt with the slow belt at 0.9 m/s and the fast belt at 1.8 m/s (2:1 ratio of fast:slow) for 6 min. Post-stroke patients performed the task twice. First, the non-paretic leg was assigned to the fast belt during split-belt conditions (denoted "paretic leg slow"). Second, the paretic leg was assigned to the fast belt (denoted "paretic leg fast"). Post-stroke patients performed one trial in each session. For control participants, the fast and slow belts were randomly assigned to either the right or left leg.

2.4. Data collection

A motion-capture system involving 22 cameras (Vicon Motion Systems, Oxford, UK, Figure 1B) was used to capture three-dimensional marker data and calculate the CoM during the task. Marker placement followed the Plug-in-Gait full-body AI model. The ground reaction force (GRF) exerted on the instrumented double-belt treadmill (Bertec Corp., OH, USA) and belt speeds were recorded bilaterally (1,000 Hz) and resampled to match the kinematic data (100 Hz). Recording of kinematic and kinetic signals was conducted for 30 s at the end (150–180 s) of the tied-belt period, and at the beginning

(180–210 s, defined as “early-adaptation”) and end (510–540 s, defined as “late-adaptation”) of the split-belt period. We calculated each CoP from the recorded GRFs and moments.

2.5. Data analysis

2.5.1 Spatiotemporal parameters and center of mass-center of pressure angle

We calculated the following spatiotemporal gait parameters: 1) Percent stance time, defined as the duration from foot contact to toe off, expressed as a percentage of stride time; 2) Step length, defined as the anteroposterior distance between the ankle markers of each leg at foot contact; 3) Percent double-support time, defined as the duration from foot contact on one side to toe off on the other side, expressed as a percentage of stride time. The angle between a vertical line (against-earth vertical) and the vector from the CoM to the CoP of the leading leg on a sagittal plane at foot contact on the slow and fast sides was calculated (θ, Figure 1C). Based on a previous study [Malone et al, 2014], we calculated the symmetrical indices of slow- and fast-side data for all parameters using Equation 1:

$$Symmetrical\ index = \frac{fast\ side\ -\ slow\ side}{fast\ side\ +\ slow\ side} \times 100(\%) \qquad (1)$$

2.5.2 Kinetic parameters

The tangent of the CoM-CoP angle is strongly correlated with the peak braking force of the leading leg (required coefficient of friction, RCOF) during walking without instability [Yamaguchi et al, 2016]. Using bipedal inverted pendulum modeling with the CoM and CoP in the sagittal plane (Figure 1C), Yamaguchi et al. estimated the CoM moment (Eq. 2):

$$Mx = Fz1(y_{COP1} - y_{COM}) + Fz2(y_{COP2} - y_{COM}) - Fy1 * Z_{COM} + Fy2 * Z_{COM} \quad (2)$$

From this, the braking force for the leading leg (Fy1/Fz1) can be defined by Equations 3 and 4:

$$\frac{Fy1}{Fz1} = tan\theta + \frac{Fz2}{Fz1} * \frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} + \frac{Mx}{Fz1 * Z_{COM}} \quad (3)$$

$$RT = \frac{Fz2}{Fz1} * \frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} + \frac{Mx}{Fz1 * Z_{COM}} \quad (4)$$

Equations 3 and 4 separate the tangent of CoM-CoP angle of the leading leg (tanθ) and terms relating to the effect of the trailing leg (residual term [RT]). Yamaguchi et al. clarified that the peak braking force of the leading leg (Fy1/Fz1) is approximately equal to tanθ at the time of peak braking force on the leading leg [Yamaguchi et al. 2016].

We used these kinetic parameters to explain the differences in the dynamic stability of post-stroke patients who either did or did not show adaptation of spatiotemporal parameters. We assessed the difference between the braking force for the leading leg (Fy1/Fz1) and tanθ during walking adaptation.

2.6. Subgroup classification for post-stroke patients

Post-stroke patients were classified as "responders" or "non-responders" by the change in CoM-CoP angle symmetry during late-adaptation. We determined the threshold for classification as a responder by the mean symmetrical index during late-adaptation in elderly subjects.

2.7. Statistics

We compared each parameter of the last 10 strides of the tied-belt and late-adaptation periods and the first 10 strides of the early-adaptation period of each treadmill session. The symmetrical index of spatiotemporal parameters normalized to tied-belt conditions (i.e., early- and late-adaptation minus the tied-belt period) was analyzed to evaluate the change from the tied-belt period. Within-subjects one-way repeated measures analysis of variance (ANOVA) was used to compare the symmetry of spatiotemporal parameters among the different periods (tied-belt, early-adaptation, and late-adaptation). Two-way repeated measures ANOVA was used to identify statistically significant interactions in the normalized index between periods and groups. If ANOVA revealed significant main effects or interactions, Bonferroni's post-hoc comparisons were carried out to identify significant differences among variables. Paired t-tests were used to evaluate the difference between the peak braking force of the leading leg and tanθ during late-adaptation within subgroups. To evaluate the difference in the symmetry of CoM-CoP angle during late-adaptation between subgroups, we used non-paired t-tests among subgroups. All analyses were performed with a significance level set at $p = 0.05$, and using MATLAB (MathWorks Inc., MA, USA).

3. Results

3.1. Center of mass-center of pressure angle and classification of post-stroke patients

Table 1 shows the characteristics of post-stroke patients. The mean symmetrical index of CoM-CoP angle during late-adaptation was 1.13 in young and 0.78 in elderly control participants. Table 2 details the recorded values for each parameter in the post-stroke subgroups, determined using the threshold of 0.78. A significant difference was seen in the symmetry of the CoM-CoP angle between subgroups ($p < 0.05$), although no other parameters or functions were significantly different.

Figure 2A illustrates the changes in the CoM-CoP angle observed for each side during the testing period at foot contact for responders, non-responders, and control participants. Responder and control subjects established symmetry of the CoM-CoP angle through split-belt walking, while non-responders remained asymmetrical. Figure 2B is a representative example of the CoP profile relative to CoM position in the horizontal plane during late-adaptation. At foot contact, the foot placement of responders was symmetrical, while that of non-responders was asymmetrical.

3.2. Spatiotemporal parameters

Figure 3 shows the changes in the symmetrical indices of stance time, step length, and double-support time at different time points during the testing period. For all participants, the symmetrical index for stance time was significantly different during early- and late-adaptation to those recorded in the tied-belt period ($p < 0.05$). Stance time remained asymmetrical in the spit-belt period. In the responder and elderly groups, the symmetrical index of step length in early-adaptation showed significant differences to the tied-belt values ($p < 0.001$). No significant difference in this parameter was seen between early and late-adaptation ($p = 0.53$). Symmetry of step length was re-established during split-belt walking in the responder and elderly groups, and there were no interactions between the period and group (F [2,42] = 0.17, $p = 0.85$, and F [2,36] = 0.15, $p = 0.86$). Similarly, symmetry of the double-support time was also re-established through spit-belt walking in the non-responder and elderly groups. Double-support times of these two groups indicated that there were no interactions between the period and group (F [2,42] = 0.32, p = 0.73, and F [2,36] = 0.36, p = 0.51). On the other hand, the step length and double-support time of non-responders were significantly different in late-adaptation compared with the tied-belt period (p < 0.01). Step lengths of the non-responder and elderly groups indicated that there were significant interactions between the period and group (F [2,66] = 5.67, p = 0.01, and F [2,72] = 3.73, p = 0.03). The stance time remained asymmetrical in the non-responder group; however, the double support time of the non-responder and elderly groups indicated that there were no interactions between the period and group (F [2,66] = 2.40, p = 0.09, and F [2,72] = 2.50, p = 0.09).

3.3. Kinetic parameters

Figure 4 shows the difference between Fy1/Fz1 and $\tan\theta$ during late-adaptation for post-stroke patients. No significant differences were observed between the fast and slow sides in the responder (paretic leg slow p = 0.30 and 0.65, paretic leg fast p = 0.11 and 0.13, respectively). In the non-responder group, there was a significant difference between the fast and slow sides in the paretic leg fast setting (p < 0.05). In the non-responder group, there was no significant difference between the values for the fast and slow sides in the paretic leg slow setting ($p = 0.05$). All three terms in the calculation of RT (Eq. 4) relate to the effect of the trailing leg. Figure 5 shows the relationship between the first and second terms on the slow and fast sides. The young adult group showed a strong negative correlation between these terms (figure 5A, $R = -0.97$; $p < 0.001$), as did the elderly and post-stroke groups (figure 5B–D, $R = -0.78$–-0.72; $p < 0.01$). Thus, the first and second terms almost cancel each other out.

4. Discussion

4.1. Difference in adaptability of gait parameters between post-stroke subgroups

Classification by symmetry of the CoM-CoP angle during split-belt treadmill walking meant that adaptabilities of spatiotemporal parameters differed between subgroups. For step length and double-support time, the interlimb parameters of non-responders who did not re-established symmetry of the CoM-CoP angle remained asymmetrical, as did the intralimb parameters. On the other hand, post-stroke patients who re-established symmetry of the CoM-CoP angle through split-belt walking also re-established baseline symmetry of interlimb parameters, as did the young and matched control participants. In healthy humans, symmetry of interlimb parameters is gradually re-established by trial-and-error during split-belt treadmill walking [Reisman et al, 2005].

However, responders did not completely re-establish symmetry of interlimb parameters, in line with a previous study [Reisman et al, 2013]. Our results support those of a previous study that suggested an asymmetric walking pattern could be the optimal pattern for certain individuals, as a byproduct of asymmetries in neural control or limb mechanics [Finley et al, 2017]. Our results clarify that complete re-establishment of CoM-CoP angle symmetry through split-belt walking depends on the adaptability of spatiotemporal parameters in individual post-stroke patients. Analyses of GRF [Ogawa et al, 2014] and the phase of gait cycle [Fujiki et al, 2015] have identified foot contact as an important element in adaptation in split-belt walking. Therefore, differences in spatiotemporal parameters suggest that the position of foot contact represents an important kinetic contribution to adaptation.

4.2. Kinetic factors in the symmetry of center of mass-center of pressure angle

Bipedal inverted pendulum modeling with the CoM and CoP revealed that the coincidence between the CoM-CoP angle (i.e., $\tan\theta$) and the peak vector of braking force (i.e., $Fy1/Fz1$) is related to the difference in CoM-CoP angle symmetry in post-stroke patients. The peak braking force of the leading leg, calculated from the vertical and anteroposterior components of the GRF, usually occurs shortly after ground contact by the leading foot. This is an important factor with regards to slipping during walking, as it must not exceed the RCOF [Nagano et al, 2013]. Peak braking forces of the leading leg during straight and turning gaits are strongly positively correlated with tanθ [Yamaguchi et al, 2013; Yamaguchi et al, 2016]. In the responder group, tanθ and Fy1/Fz1 were consistent, as is observed in healthy people, whereas non-responders showed significant differences between tanθ and Fy1/Fz1. However, the first and second terms of the RT almost canceled each other in all participants, indicating that the RT directly reflects the third term in Eq. 4, which defines the moment around the CoM. Therefore, in non-responders whose RT was not 0, the moment around the CoM from the leading and trailing legs was not canceled at the time of RCOF. The difference between tanθ and the RCOF is increased when dynamic stability is decreased; for example, during slow walking [Yamaguchi et al, 2017]. Therefore, post-stroke patients who do not re-establish symmetry of the CoM-CoP angle might experience low dynamic stability in split-belt walking.

Bipedal locomotion is governed primarily by passive dynamics with minimal active energy costs [Collins et al, 2005]. Foot contact of the leading leg represents negative work, which produces a backward rotational moment around the CoM in the sagittal plane from the leading leg. In an ideal dynamic walking model, this negative work can be reduced by pushing off with the trailing leg just before foot contact [Kuo, 2002]. Kuo et al indicate that the trailing leg reduces the amount of active work that is required to predictively sustain steady gait [Kuo, 2010]. Therefore; at foot contact, the interaction between the leading and trailing legs (interlimb coordination) is also key in stabilization of the whole body.

4.3. Limitations and future study

This study had some limitations. First, our protocol did not include a tied-belt condition in the post-adaptation period. Assessments of the parameters after split-belt walking could provide evidence of motor learning. However, as our main goal was to provide information about the changes during adaptation in split-belt walking, this was not carried out. Therefore, we analyzed adaptive behavior through split-belt treadmill walking and do not discuss motor learning in this study.

Second, we allowed patients to use the handrail for support and safety. This has been suggested to affect symmetry in post-stroke patients [IJmker et al, 2015; Finley et al, 2017]. However, the activity levels of our post-stroke patients were, in general, lower than those of previous studies. The novelty of this study was the assessment of differences in patients with low levels of activity.

Third, we did not observe a statistically significant effect of split-belt walking in patients of the non-responder groups. This was particularly notable in the case of double-support time, and suggests that the classification of subjects may not always have been correct. Future studies including larger sample sizes are required.

4.4. Conclusions

Our study shows that post-stroke patients who re-establish symmetry of the CoM-CoP angle at foot contact can adapt their walking for split-belt perturbation. This suggests that predictively symmetric foot placements on the fast and slow sides are necessary for adaptation in walking. Reisman et al argue that adapting the angle of the leading leg at weight transfer is critical to control the relative position between the CoM and the leading leg. Symmetrical foot placement is achieved by interlimb coordination of both legs, and may contribute to dynamic stability. Because patients who have no limitations in forward placement of the paretic leg are able to use the split-belt treadmill, this may be one of the criteria for split-belt training in post-stroke patients.

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17

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Figure captions

Fig. 1 (A) Illustration of the split-belt walking protocol. (B) The motion capture system and split-belt treadmill. Kinematic and trajectory data as well as belt speed and forces data were synchronized. (C) The bipedal inverted pendulum model. Abbreviations: CoM, whole-body center of mass; yCoM (zCoM), anterioposterior (vertical) position of the CoM; Mx, moment around the CoM; Fy1/Fz1 (Fy2/Fz2), braking force for the leading (trailing) leg; CoP1 (CoP2), center of pressure of the leading (trailing) leg; yCoP1 (yCoP2), anterioposterior position of the CoP of the leading (trailing) leg; θ, the CoM-CoP angle of the leading leg.

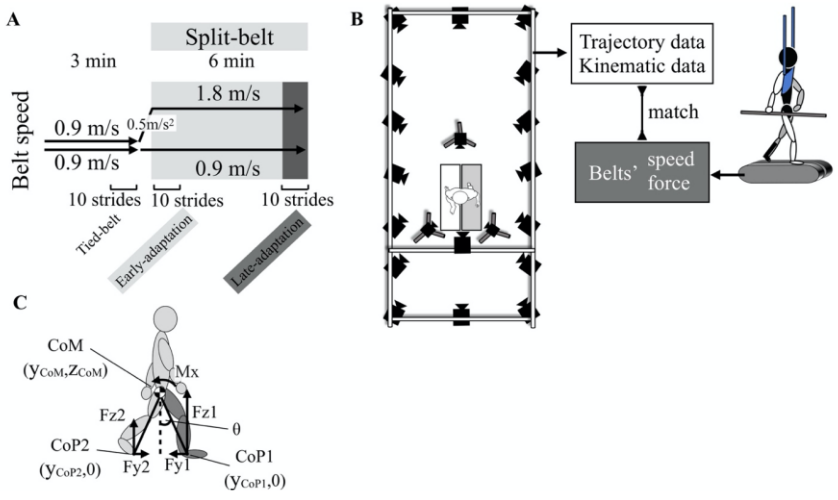
Fig. 2 (A) Symmetrical index of center-of-mass (CoM)-center-of-pressure (CoP) angle measurements for sequential strides from representative young (top row), elderly (second row), and post-stroke patients (third row, responder; bottom row, non-responder) across all testing periods. (B) The CoP profile relative to CoM for the same representative subjects as in (A); representative young (top left), elderly (top right), and post-stroke patients (bottom left, responder; bottom right, non-responder) during the same cycle of late-adaptation. Definitions: CoM, centre of mass; CoPx, mediolateral position of the CoP; CoPy, anterioposterior position of the CoP.

Fig. 3 The symmetrical index of (A) Stance time, (B) step length, and (C) double-support time measurements for sequential strides from representative young (top row), elderly (second row), and post-stroke patients (third row, responder; bottom row, non-responder) across all testing periods. (D) Stance time, (E) step length, and (F) double-support time differences for post-stroke (red and blue circles represent responders with paretic leg slow and fast settings, respectively; red and blue squares represent non-responders with paretic leg slow and fast settings, respectively), young (triangle), and elderly (diamond) groups. Each data point indicates the difference from the tied-belt period (early- or late-adaptation minus the tied-belt data), averaged across all subjects in the group. Error bars indicate mean standard error. Filled objects indicate statistically significant differences compared with tied-belt values.

Fig. 4 Comparisons between required coefficient of friction (gray) and tanθ (black) values of responders and non-responders for fast and slow belt sides in (A) paretic leg slow and (B) paretic leg fast settings. Error bars indicate mean standard deviation. Horizontal bars indicate significant differences ($p < 0.05$). Abbreviations: RCOF, required coefficient of friction.

Fig. 5 Graphs of the negative relationship between the first and second terms of RCOF in Eq. (3) for (A) young, (B) elderly, (C) post-stroke patients in paretic leg slow settings, and (D) post-stroke patients in paretic leg fast settings. In C and D, red and blue dots represent data from responders and non-responders, respectively.

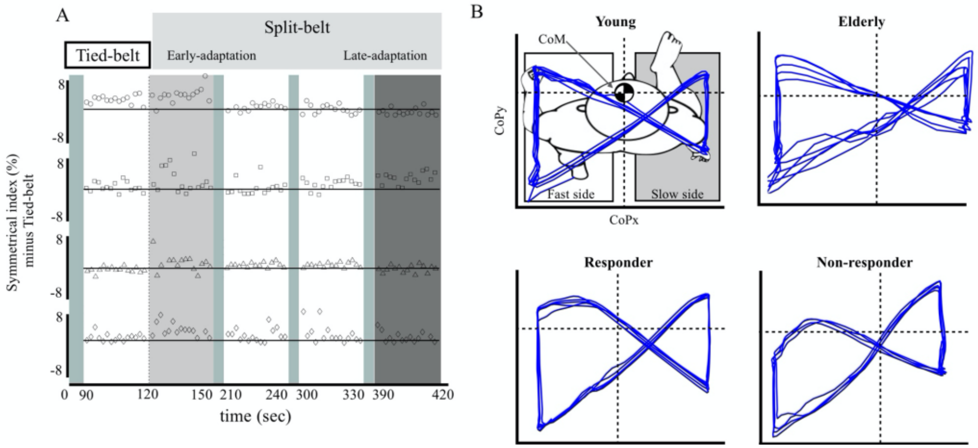
Figure 1



A
Split-belt
3 min
6 min
Belt speed
1.8 m/s
0.9 m/s
0.5m/s²
0.9 m/s
0.9 m/s
10 strides
10 strides
10 strides
Tied-belt
Early-adaptation
Late-adaptation
B
Trajectory data
Kinematic data
match
Belts' speed force
C
CoM
(y_{CoM},z_{CoM})
Mx
Fz2
Fz1
θ
CoP2
(y_{CoP2},0)
Fy2
Fy1
CoP1
(y_{CoP1},0)

1

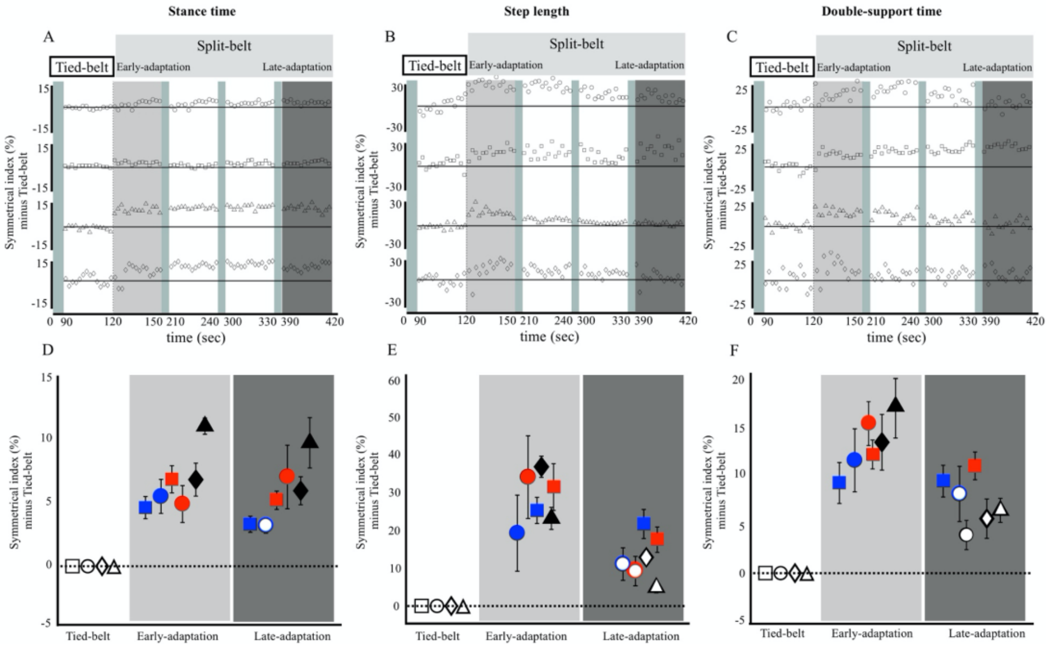
Figure 2



A
Split-belt
Tied-belt
Early-adaptation
Late-adaptation
Symmetrical index (%) minus Tied-belt
8
-8
0 90 120 150 210 240 300 330 390 420
time (sec)
B
Young
CoM
CoPy
CoPx
Fast side
Slow side
Elderly
Responder
Non-responder

1

Figure 3

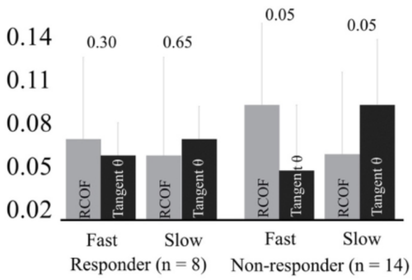


Stance time
Step length
Double-support time
A
B
C
Split-belt
Tied-belt
Early-adaptation
Late-adaptation
Symmetrical index (%) minus Tied-belt
time (sec)
D
E
F
Tied-belt
Early-adaptation
Late-adaptation

1

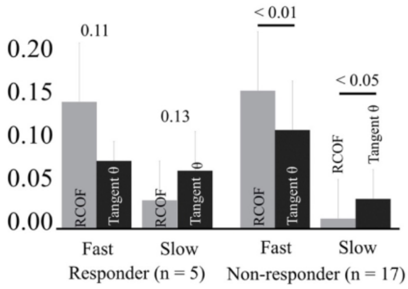
Figure 4

A



0.14
0.11
0.08
0.05
0.02
0.30
0.65
0.05
0.05
RCOF
Tangent θ
Fast
Slow
Responder (n = 8)
Non-responder (n = 14)

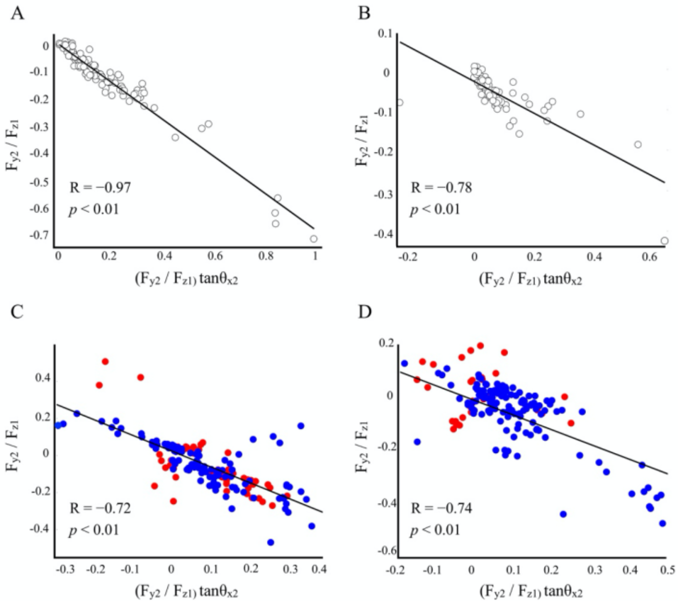
B



0.20
0.15
0.10
0.05
0.00
0.11
0.13
< 0.01
< 0.05
RCOF
Tangent θ
Fast
Slow
Responder (n = 5)
Non-responder (n = 17)

1

Figure 5



A
B
C
D
R = −0.97
p < 0.01
R = −0.78
p < 0.01
R = −0.72
p < 0.01
R = −0.74
p < 0.01
Fy2 / Fz1
(Fy2 / Fz1) tanθx2

1

Table 1. Post-stroke patient characteristics (n = 22).

Subject	Age	Sex	Hemiparetic side	LE FMA (/34)	Months since stroke	Speed slow/fast (m/s)	FIM	TUG (s)
S1	68	M	R	21	128	0.25/0.5	115	17.80
S2	63	F	L	20	27	0.3/0.6	115	31.10
S3	69	M	R	29	142	0.45/0.9	126	7.00
S4	75	M	R	22	120	0.35/0.7	115	15.80
S5	70	F	R	21	65	0.3/0.6	120	14.00
S6	79	M	L	31	13	0.2/0.4	105	12.00
S7	73	M	L	26	36	0.25/0.5	120	16.00
S8	78	M	R	34	9	0.4/0.8	126	10.40
S9	65	M	L	25	7	0.35/0.7	120	13.97
S10	71	M	R	20	18	0.2/0.4	114	18.80
S11	73	M	R	22	8	0.2/0.4	107	25.90
S12	70	M	L	19	141	0.2/0.4	104	45.12
S13	57	M	L	19	18	0.2/0.4	118	30.30
S14	45	F	L	21	66	0.2/0.4	114	15.80
S15	48	F	L	14	132	0.2/0.4	119	28.59
S16	84	M	R	29	7	0.2/0.4	121	24.80
S17	71	M	R	27	162	0.2/0.4	109	21.81
S18	70	M	R	20	33	0.2/0.4	106	32.50
S19	63	M	L	28	102	0.2/0.4	83	46.88
S20	65	F	L	28	47	0.2/0.4	121	21.82
S21	69	M	R	25	50	0.2/0.4	102	27.00
S22	55	M	L	14	103	0.2/0.4	116	37.65

Abbreviations: F, female; FIM, functional independence measure; M, male; L, left; LE FMA, lower extremity Fugl-Meyer Assessment; R, right; TUG, Timed Up and Go test.

Table 2. Comparisons of parameters between responders and non-responders for paretic leg fast and slow settings.

	Paretic leg slow			Paretic leg fast		
	Responder (n = 8)	Non-responder (n = 14)	*p* value	Responder (n = 5)	Non-responder (n = 17)	*p* value
LE FMA (/34)	22.2 (7.0)	23.8 (4.6)	0.52	21.2 (7.1)	23.9 (5.0)	0.35
TUG (s)	21.9 (10.7)	24.3 (11.1)	0.63	22.0 (10.1)	23.8 (11.2)	0.75
FIM (/124)	114.7 (7.8)	112.7 (10.8)	0.64	111.4 (5.0)	114.0 (10.7)	0.60
Maximum walking speed (m/s)	0.5 (0.2)	0.5 (0.1)	0.38	0.5 (0.1)	0.5 (0.1)	0.59
Symmetrical index of percent stance time during tied-belt period	7.9 (5.5)	6.3 (5.9)	0.54	8.4 (4.1)	7.0 (5.3)	0.49
Symmetrical index of step length during tied-belt period	32.7 (30.4)	36.3 (29.0)	0.70	50.9 (33.8)	31.7 (29.9)	0.44
Symmetrical index of percent double-support time during tied-belt period	18.9 (12.1)	13.6 (11.8)	0.16	16.2 (15.0)	15.1 (9.4)	0.18
Symmetrical index of CoM-CoP angle during late-adaptation period	0.28 (0.20)	2.43 (2.09)	<0.01	0.43 (0.21)	2.73 (2.19)	<0.05

Data are presented as number (%). Abbreviations: CoM, center of mass; CoP, center of pressure; RIM, functional independence measure; LE FMA, lower extremity Fugl-Meyer Assessment; TUG, Timed Up and Go test.

1

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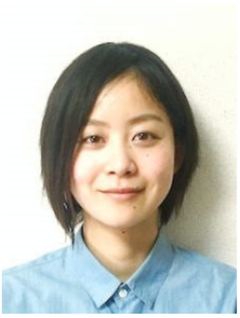

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