2010270013A

厚生労働科学研究費補助金

障害者対策総合研究事業（身体・知的等障害分野）

脊髄損傷後の歩行機能回復のための新たなニューロリハビリテーション方法の開発に関する研究

平成22年度　総括・分担研究報告書

研究代表者　赤居　正美

平成23（2011）年４月

目　　次

厚生労働科学研究費補助金（障害者対策総合研究事業（身体・知的等障害分野））
総括研究報告書

脊髄損傷後の歩行機能回復のための新たなニューロリハビリテーション方法の開発に関する研究

研究代表者　　赤居　正美　　国立障害者リハビリテーションセンター病院　院長

研究要旨

本研究では、脊髄損傷患者の歩行機能再獲得の鍵と目される脊髄パターン発生器（central pattern generator: CPG）の性質に着目し、その活動を励起・賦活させる神経生理学的機序を検証する。それを基に新たな神経リハビリテーション（neurorehabilitation）方法の構築をめざす。
近年、特に欧米を中心に、体重を部分的に免荷した状況下で下肢の動作を補助し、正常な歩容を再現する「免荷式歩行トレーニング」が、歩行リハビリテーションの主流となりつつある。この動向には、近年の神経科学領域の研究によって脊髄組織を含む中枢神経系が可塑的性質を持つことが明らかにされ、繰り返しの歩行訓練によって,かなりの歩行機能再獲得が実現できる可能性が示されたことが影響している。
最近報告された無作為化比較試験では、ASIA CおよびDに分類される脊髄不全損傷者のうち、９割にトレーニングによる歩行能力の改善が認められている。しかし従来の歩行リハビリテーションとの比較では、その効果に統計学的有意差は認められず、どのような戦略を採ればより効果的な歩行機能再獲得が実現できるのか、あるいは障害や麻痺状態などによってどの程度までの機能回復を見込めるのか、については未だ不明な部分が多い。また、免荷式歩行トレーニングの効果を検証した幾つかの研究では、介入後の歩行速度や動作の改善、運動中の筋活動の改善などの効果を報告しているものの、歩行機能改善の具体的な神経機序については、明らかにされていない。
具体的には、１年目に脊髄損傷者の残存神経機能の定量的把握と、歩行運動出力を高めるための具体的な方策を検討し、その成果に基づいて、２年目に歩行リハビリテーションプログラムを作成する。最終年度には、開発したリハビリプログラムの評価・改善を行うとともに、歩行リハビリテーションによる健康・体力、ADL、社会生活への波及的効果についても包括的に検討する。
本研究の成果は、脊髄損傷者の移動能力の維持・向上を図る、より効果的な新たな神経リハビリテーション方法を立案する上で極めて有用な情報を提供し得るものと考えられる。

研究分担者氏名・所属研究機関名及び所属研究機関における職名
緒方　徹
（国立障害者リハビリテーションセンター研究所　部長）
中澤　公孝
（東京大学大学院総合文化研究科　教授）
飛松　好子
（国立障害者リハビリテーションセンター病院　部長）
神作　憲司
（国立障害者リハビリテーションセンター研究所　室長）
梅崎　多美
（国立障害者リハビリテーションセンター学院　教官）

A．研究目的

体重免荷によるトレッドミル歩行は「正常な歩行動作を再現することにより種々の求心性感覚入力を脊髄CPGに与え、その活動を改善する」という理論的基盤をもつ。しかし、実際のところ歩行機能の再獲得に至る神経生理学的機序は未だ明らかではない。本研究では、歩行機能再獲得の鍵と目される、脊髄CPGの性質に着目し、その活動を励起させる神経生理学的機序を検証し、それを基にした新たな神経リハビリテーション方法の提案を目指す。
すなわち、本研究の目的は脊髄CPGの活動を励起・賦活する適切な神経入力を与えることにより可塑的変化を促し、合目的的に歩行機能回復を実現する神経リハビリテーション方法を開発することである。

B．研究方法

（１）脊髄CPGを賦活化するというコンセプトはあるものの、それ以上のメカニズムが明らかでないこともあり、訓練方法（Lokomatの使用方法も含め）にばらつきがある。その背景として、受傷後1年以内の訓練効果には自然回復の影響があり、他の訓練効果と識別しにくいこと、逆に受傷後1年以降であると訓練効果が出にくいことも考えられる。したがって、研究の組み立てとして、以下を定めた。
・CPG賦活化の方法を検討する。
・訓練前後で得られる歩行の変化を詳細に解析する。
・まず受傷後1年以上経過した症例を対象とする。
そこで、
１年目を脊髄CPGの活動賦活のための方法論づくりに関する基礎研究ステージ、２年目をリハビリテー

ション方法の開発に関する応用研究ステージ、さらに３年目を開発した方法の評価・改良に関する最終研究ステージとして位置づける。

（３）応用研究ステージにあたる平成22年度には、1年目に検討した、脊髄CPGの特性についての知見を応用し、歩行運動出力を適切に促すための具体的な方法の検索と、考案された方法を実際に脊髄不全損傷者に実施することで、歩行機能がどのように変化するかを実験的に検討した。

（４）具体的には、動力型歩行補助装置（Lokomat）による外的な歩行キネマティクスの形成を軸として、歩行運動出力を促通すると考えられる種々の末梢性感覚情報、異なる体肢からの神経情報、付加的な電気刺激による感覚入力、脳からの重畳的な随意神経指令を組みあわせた新たな方法を考案、3ヶ月の歩行リハビリテーション実施による訓練効果の定量的把握を行った（図1）。

（５）合わせて脊髄における組織破壊に由来する微量物質（バイオマーカー）を指標に組織損傷の程度を推測し、臨床的機能予後との関連を検討した。

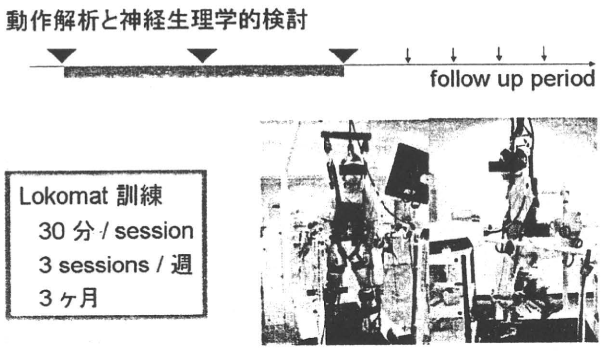


図1 Lokomatを用いた歩行訓練の実際

（倫理面への配慮）

研究は国立障害者リハビリテーションセンター倫理審査委員会の承認を得た上で実施している（取得済）

C．研究結果

昨年度（初年目）は、Lokomatを用いて脊髄不全損傷（不全対麻痺）患者のトレーニング実験を行い、介入効果の評価法としての経頭蓋磁気刺激(transcranial magnetic stimulation; TMS)と下肢筋の誘発運動電位(motor evoked potential; MEP)、とりわけ前脛骨筋との関連性を見ることが出来た。また足底を中心とした荷重情報の有無や上肢運動の付加、被験者の集中力などの、いいかえればactive assistive trainingの重要性が証明された。

本年度（二年目）は初年度の結果を踏まえ、Lokomatを用いた歩行訓練を施行する際に、適度な荷重・上肢運動の追加・随意的な歩行努力の3要素を加えたプロトコールを作成した。実際に不全脊髄損傷患者に対し、週3回のLokomatトレーニングを12週間行い、その効果を前後の測定によって検証することとした。12週間のトレーニング実験を22年度中に5名終了する予定であったが、このたびの震災により2名が1ヶ月の期間延長となった。詳細な結果については未だ解析途中であるが、いずれの被験者においても歩行動作の改善、痙性麻痺の減少傾向が認められた。皮質脊髄路興奮性を反映する運動誘発電位は、3名中2名においてトレーニング当初は発現しなかった前脛骨筋の応答がトレーニング後で発現するなど、中枢神経の可塑的変化を支持する結果が得られた。また、歩行中の下肢筋活動に関して、体重支持・抗重力的に働く大腿直筋の活動がトレーニング経過に伴って増加する傾向、さらに伸張反射感受性の過剰亢進によって生じると思われる遊脚期後半の一過性の大腿二頭筋の活動がトレーニング後に減弱する傾向が認められた。

D．考察

これらの結果は、繰り返しの求心性感覚情報による脊髄歩行中枢の活動喚起と、歩行運動に必要な随意運動出力増加によってもたらされるものと考えられる。

本研究を通して歩行機能再獲得の神経機序の解明と、効果的に歩行再獲得を促す方法が開発されれば、多くの運動機能障害者が効果的な歩行リハビリテーションを行う環境を整備することが可能となるものと考えられる。これらの研究を通して、脊髄損傷者の歩行機能再獲得のために具体的な方法論の提示が可能となろう。

E．結論

12週間のトレーニング実験を最終的に5名終了する予定である。残りのトレーニング実験結果の集約を図るとともに、トレーニング実験を進める過程で被験者の歩行特性に応じた装置の動作設定の見直しやトレーニングプロトコルのアップデートなど、適宜評価・改善を行う。

F．健康危険情報

特になし

G．研究発表

1.　論文発表

- Kamibayashi K, Nakajima T, Fujita M, Takahashi M, Ogawa T, Akai M, Nakazawa K. Effect of sensory inputs on the soleus H-reflex amplitude during robotic passive stepping in humans. Experimental Brain Research, 202:385-395, 2010.
- Abe MO, Masani K, Nozaki D, Akai M, Nakazawa K. Temporal correlations in center of body mass fluctuations during standing and wal

king. Human Movement Science, 29:556-566, 2010.

- Sayenko DG, Alekhina MI, Masani K, Vette AH, Obata H, Popovic MR, Nakazawa K. Positive effect of balance training with visual feedback on standing balance abilities in people with incomplete spinal cord injury. Spinal Cord, 48:886-893, 2010.

2. 学会発表
（発表誌名巻号・頁・発行年等も記入）

H. 知的財産権の出願・登録状況
（予定を含む。）
1. 特許取得
無
2. 実用新案登録
無
3. その他

厚生労働科学研究費補助金（障害者対策総合研究事業（身体・知的等障害分野））
分担研究報告書

脊髄損傷後歩行機能再獲得のための神経リハビリテーション方法の開発

研究分担者　　緒方　徹　　国立障害者リハビリテーションセンター研究所　部長

研究要旨
近年、神経科学領域で注目されている脊髄可塑性、脊髄CPGの知見を基盤とし、科学的根拠に立脚したあらたな神経リハビリテーション（neurorehabilitation）方法の構築を目指す。本研究では、脊髄における組織破壊に由来する微量物質（バイオマーカー）を指標に組織損傷の程度を推測し、臨床的機能予後との関連を検討する。

A．研究目的

体重免荷によるトレッドミル歩行は「正常な歩行動作を再現することにより種々の求心性感覚入力を脊髄CPGに与え、その活動を改善する」という理論的基盤をもつ。しかし、実際のところ歩行機能の再獲得に至る神経生理学的機序は未だ明らかではない。それ以上のメカニズムが明らかでないことと共に、訓練方法（Lokomatの使用方法も含め）にもばらつきがある。その背景として、受傷後1年以内の訓練効果には自然回復の影響があり、他の訓練効果と識別しにくいこと、逆に受傷後1年以降であると訓練効果が出にくいことも考えられる。したがって、ここでは自然回復の関与分を明らかにするべく、定量的な機能的予後推測法を開発する。

B．研究方法

血液などの検体に含まれる特定のたんぱく質を定量解析することで疾病の状態を評価することは広く行われている。これまで脊髄損傷に対しその重症度を評価するための指標（バイオマーカー）には確立したものはない。近年、脊髄の組織破壊に伴って血液中に放出される神経軸索構成蛋白pNF-Hが血液中で測定可能であることが報告された。そこで、このpNF-Hの血中濃度が脊髄損傷の重症度を反映し、自然経過の予後推定に役立つかを検証した。

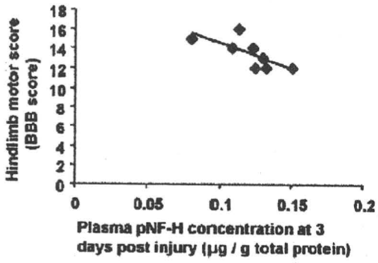
まず、ラットを用いた脊髄損傷モデルにおいて、血液中のpNF-HをELISA法によって測定、最終的な脊髄運動機能に相当する受傷4週後の後肢運動機能との相関の有無を検討した。次いで、ヒト脊髄損傷症例においてのpNF-H血中濃度変化を臨床例において確認した。なお、採血にあたっては国立障害者リハビリテーションセンターおよび血液サンプルを採取した病院での倫理委員会の承認を得て行った。

C．研究結果

脊髄損傷から3日目に血中pNF-Hは最高値を示し、その値は最終的な運動機能と統計学的に有意な相関を示すことがわかった（図1）。

図1

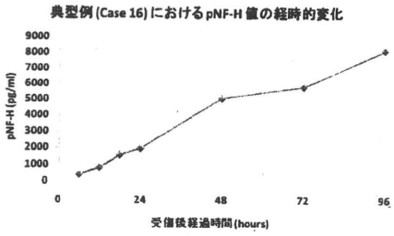
頸椎



血中pNF-H

濃度の推移を測定したところ、単調増加の変化パターンを認めた(図2)。

図2



D．考察

血中でのpNF-H高値はより多くの組織損傷を示唆し、その結果脊髄機能である後肢運動機能が低下していると考えられる。また、ラットと異なりヒト脊髄損傷ではpNF-Hは受傷後3日目に最高値を示すのではなく、その後も漸増することが明らかとなった。

E．結論

今後、受傷後のpNF-H濃度がヒト脊髄損傷例の予後推定に有用であるかを検討するうえで、貴重な資料となるものと考えられる。

F．健康危険情報

特になし

G．研究発表

1.　論文発表

Ueno T, Ohori Y, Ito J, Hoshikawa S, Yamamoto S, Nakamura K, Tanaka S, Akai M, Tobimatsu Y, Ogata T:Hyperphosphorylated neurofilament NF-H as a biomarker of the efficacy of minocycline therapy for spinal cord injury. Spinal Cord 49(3):333-336, 2011

2．発表

特になし

F．知的財産権の出願・登録状況

1.　特許取得

特になし

2.　実用新案登録

特になし

3.　その他

研究成果の刊行に関する一覧表

書籍

著者氏名	論文タイトル名	書籍全体の編集者名	書　籍　名	出版社名	出版地	出版年	ページ
		中澤　公孝	歩行のニューロリハビリテーション—歩行の再獲得をめざした理論と臨床	杏林書院	東京	2010	167

雑誌

発表者氏名	論文タイトル名	発表誌名	巻号	ページ	出版年
Kamibayashi K, Nakajima T, Fujita M, Takahashi M, Ogawa T, Akai M, Nakazawa K.	Effect of sensory inputs on the soleus H-reflex amplitude during robotic passive stepping in humans.	Experimental Brain Research	202	385-395	2010
Abe MO, Masani K, Nozaki D, Akai M, Nakazawa K.	Temporal correlations in center of body mass fluctuations during standing and walking	Human Movement Science	29	556-566	2010.
Sayenko DG, Alekhina MI, Masani K, Vette AH, Obata H, Popovic MR, Nakazawa K.	Positive effect of balance training with visual feedback on standing balance abilities in people with incomplete spinal cord injury	Spinal Cord	48	886-893	2010
Ueno T, Ohori Y, Ito J, Hoshikawa S, Yamamoto S, Nakamura K, Tanaka S, Akai M, Tobimatsu Y, Ogata T	Hyperphosphorylated neurofilament NF-H as a biomarker of the efficacy of minocycline therapy for spinal cord injury	Spinal Cord	49	333-336	2011

研究成果の刊行物

Exp Brain Res
DOI 10.1007/s00221-009-2145-2

RESEARCH ARTICLE

Effect of sensory inputs on the soleus H-reflex amplitude during robotic passive stepping in humans

Kiyotaka Kamibayashi · Tsuyoshi Nakajima · Masako Fujita · Makoto Takahashi · Tetsuya Ogawa · Masami Akai · Kimitaka Nakazawa

Received: 3 December 2008 / Accepted: 15 December 2009

Abstract We investigated the modulation of the soleus (Sol) Hoffmann (H-) reflex excitability by peripheral sensory inputs during passive stepping using a robotic-driven gait orthosis in healthy subjects and spinal cord-injured patients. The Sol H-reflex was evoked at standing and at six phases during passive stepping in 40 and 100% body weight unloaded conditions. The Sol H-reflex excitability was significantly inhibited during passive stepping when compared with standing posture at each unloaded condition. During passive stepping, the H-reflex amplitude was significantly smaller in the early- and mid-swing phases than in the stance phase, which was similar to the modulation pattern previously reported for normal walking. No significant differences were observed in the H-reflex amplitude between the two unloaded conditions during passive stepping. The reflex depression observed at the early part of the swing phase during passive stepping might be attributed to the sensory inputs elicited by flexion of the hip and knee joints. The present study provides evidence that peripheral sensory inputs have a significant role in phase-dependent modulation of the Sol H-reflex during walking, and that the Sol H-reflex excitability might be less affected by load-related afferents during walking.

Keywords Hoffmann reflex · Reflex modulation · Load-related afferent input · Locomotion · Spinal cord injury

K. Kamibayashi (✉)
Graduate School of Systems and Information Engineering, University of Tsukuba, 1-1-1 Tennodai, Tsukuba, Ibaraki 305-8573, Japan
e-mail: kamibayashi@iit.tsukuba.ac.jp

K. Kamibayashi · T. Nakajima · M. Takahashi · M. Akai · K. Nakazawa
Department of Rehabilitation for Movement Functions, Research Institute of National Rehabilitation Center for Persons with Disabilities, 4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan

M. Fujita
Graduate School of Engineering, Shibaura Institute of Technology, 307 Fukasaku, Minuma-ku, Saitama, Saitama 337-8570, Japan

M. Takahashi
Graduate School of Health Sciences, Hiroshima University, 1-2-3 Kasumi, Minami-ku, Hiroshima, Hiroshima 734-8551, Japan

T. Ogawa
Graduate School of Human Sciences, Waseda University, 2-579-15 Mikajima, Tokorozawa, Saitama 359-1192, Japan

Published online: 01 January 2010

Introduction

It is now generally recognized that the Hoffmann (H-) reflex, which is evoked by an electrical stimulation of group Ia afferents, is strongly modulated in a task-dependent manner between different motor tasks, and in a phase-dependent manner during rhythmic movements, such as walking, running, and pedaling in humans (Capaday and Stein 1986, 1987; Crenna and Frigo 1987; Brooke et al. 1991; Simonsen and Dyhre-Poulsen 1999; Schneider et al. 2000; Zehr 2002; Ethier et al. 2003; Stein and Thompson 2006). For example, the H-reflex excitability in the soleus (Sol) muscle is significantly lower during walking than during standing (Capaday and Stein 1986; Zehr 2002; Ethier et al. 2003). Furthermore, during walking, the Sol H-reflex amplitude increases progressively in the stance phase, while it is very small or totally absent in the swing phase (Capaday and Stein 1987; Simonsen and Dyhre-Poulsen 1999; Ethier et al. 2003).

Although modulation of reflex excitability is commonly considered to be functionally important for locomotion, neural mechanisms of reflex modulation are still not fully understood. A passive movement paradigm has often been used to investigate the mechanisms of reflex modulation during human movements (McIlroy et al. 1992; Brooke et al. 1993; Cheng et al. 1995; Misiaszek et al. 1995). Passive movement substantially reduces the influence of descending commands onto spinal motoneurons and presumably onto segmental interneurons. Therefore, changes in the H-reflex amplitude during passive movement are considered to be due to peripherally mediated sensory signals. Due to experimental difficulty in investigating the reflex modulation during walking, passive lower-limb pedaling has been substituted for passive limb movement in a walking manner (McIlroy et al. 1992; Brooke et al. 1993; Cheng et al. 1995; Misiaszek et al. 1995). The results obtained from the passive pedaling studies have revealed that the reflex amplitude in the Sol muscle substantially decreases with limb rotation, and that the degree of inhibition is dependent on the cycle phase, which is pronounced when the hip and knee joints are flexed (McIlroy et al. 1992; Cheng et al. 1995). In addition, increasing the speed of passive pedaling enhances the H-reflex depression (McIlroy et al. 1992; Cheng et al. 1995). Therefore, it has been concluded that sensory inputs from the muscle spindle have a powerful influence on the H-reflex excitability.

However, how observations from these passive "pedaling" studies can be generalized to the "walking" is still unclear. Although cyclic pedaling consists of multi-segmental movements in the lower limb, the posture of pedaling differs from the upright posture of walking. Furthermore, load-related sensory inputs during walking, which occur rhythmically through foot contact with ground, are different from those that occur during pedaling. This load-related sensory information is considered to be one of the important sensory inputs that control the locomotor activity (Harkema et al. 1997; Van de Crommert et al. 1998; Dietz and Duysens 2000; Dietz et al. 2002). In cats, it has been suggested that the afferent inputs from load receptors act on the spinal central pattern generator (Duysens and Pearson 1980; Pearson and Collins 1993; Duysens et al. 2000). In humans, the essential role of the load-related inputs to locomotor activity has also been reported in patients with spinal cord injury (SCI) (Harkema et al. 1997). Bastiaanse et al. (2000) suggested that the load receptors are involved in phasic modulation of the medium latency response of the cutaneous reflex during walking. The main receptors for detecting load information in mammals are considered to be the Golgi tendon organs and cutaneous receptors on the soles of feet (Duysens et al. 2000). Additionally, muscle spindle and joint receptors, like Ruffini endings and Pacinian corpuscles, are thought to be accessory receptors for load information. Grey et al. (2007) found that the feedback from the load receptors, especially Golgi tendon organs, contributed to the enhancement of the ankle extensor muscle activity during the late-stance phase of human walking.

Brooke et al. (1995) reported that the Sol H-reflex was attenuated over an entire passive stepping cycle manipulated by an experimenter. The study by Brooke et al. (1995), however, was performed while subjects were lying supine or tilted from the vertical position. Knikou and Conway (2001) observed that applying mechanical pressure to the foot sole inhibits the H-reflex during sitting. Therefore, it is hypothesized that load-related inputs elicited during passive stepping may also affect the Sol H-reflex excitability. On the other hand, with regard to the Ib pathway from the gastrocnemius muscle to the Sol muscle, it has been shown that, although an electrical stimulation of Ib afferents inhibited the Sol H-reflex during lying supine and sitting, the Ib inhibition was mostly absent in conditions that involved a form of loading (Faist et al. 2006).

As an alternative source for H-reflex modulation during normal walking, a central origin has been proposed. Some studies have shown that the Sol H-reflex inhibition is closely associated with the activation of the antagonistic muscle, and have suggested that the centrally produced reciprocal inhibition serves as the mechanism for H-reflex inhibition during the swing phase (Lavoie et al. 1999; Schneider et al. 2000). By investigating the H-reflex modulation during passive stepping without the tibialis anterior (TA) muscle activity, the contribution of reciprocal inhibition to the Sol H-reflex modulation might be excluded. In addition, Garrett et al. (1999) have suggested that the H-reflex modulation during walking is associated with the descending motor command that produces the stepping movement. Therefore, investigation of the H-reflex modulation in SCI patients during passive stepping might have considerable significance for clarifying the effect of the supraspinal input on the H-reflex modulation.

As driven gait orthosis (DGO) has recently been developed as a rehabilitation device for locomotor training of patients with gait disorders (Colombo et al. 2000), applying the DGO to healthy humans makes it possible to impose passive stepping. Therefore, the aim of this study was to investigate whether the Sol H-reflex is modulated in a phase-dependent manner by substantially reduced descending motor command during passive stepping using DGO in healthy subjects and SCI patients. Furthermore, this study also aimed to investigate the effects of the load-related inputs by comparing the Sol H-reflex amplitudes at different body weight unloading conditions during passive stepping.

Methods

Subjects

Ten healthy subjects (6 male and 4 female) with no history of neuromuscular disorders (22–32 years), two clinically motor-complete SCI subjects (32-year-old woman, lesion level T7, duration of injury 83 months; 19-year-old man, lesion level T12, duration of injury 6 months), and one motor-incomplete SCI subject (21-year-old male, lesion level T5, duration of injury 7 months) participated in this study. This study was conducted with an ethical approval from the local ethics committee. Each subject provided informed consent for the experimental procedures as required by the Declaration of Helsinki.

Stepping condition

Passive stepping was conducted using a DGO (Lokomat®, Hocoma AG, Switzerland, Fig. 1), a detailed description of which can be found elsewhere (Colombo et al. 2000). Briefly, DGO provides electromechanical drives for physiological hip and knee joint movements like normal walking, and imposes stepping in SCI patients and healthy subjects with substantially reduced descending command. The DGO was secured to the subject with straps across the pelvis and chest. The lower-limb parts of the orthosis were fixed to the subject with straps around the thigh and shank.

Two passive stepping conditions were performed at 1.5 km/h under different body weight unloading conditions. One was passive stepping on a treadmill with 40% unloading of body weight (ground stepping). The other stepping condition was full (100%) body weight unloading, which means that the subject was suspended with the DGO in air so that his/her feet did not touch the treadmill during passive stepping (air stepping). Body weight unloading was achieved by a parachute harness connected to counter weights (Fig. 1). Although dorsiflexion of the ankle joint during the swing phase was achieved by passive foot lifters (spring-assisted elastic straps), an ankle foot orthosis (fixed at 5° dorsiflexion) was attached to minimize movement of the ankle joint in both stepping conditions (Fig. 1). During passive stepping, the subject was instructed to relax and allow the lower-limb movements imposed by the DGO. Because body weight-supported treadmill training for SCI generally starts at a stepping speed of 1.5 km/h with 40% unloading, this stepping condition was selected from a safety standpoint for the SCI patients. In addition, the slow stepping speed allowed the healthy subjects and SCI patients to relax easily during passive stepping.

Measurements

Electromyographic (EMG) activities from the rectus femoris, biceps femoris, medial gastrocnemius, Sol, and TA muscles in the right leg were recorded with surface bar-electrodes (inter-electrode distance 10 mm) placed over the muscle bellies. The EMG signals were amplified (1,000×) and band-pass filtered (15–1,000 Hz) using a bioamplifier (MEG-108, Nihon Kohden, Japan). For measurements of the maximal voluntary EMG activities in the Sol and TA muscles, the healthy subjects performed maximum voluntary contractions of plantar flexion and dorsiflexion under the standing posture. During passive stepping, the joint angles at the hip and knee of orthosis were provided by potentiometers of the DGO. The ankle joint angle was recorded by an electrogoniometer (SG110, Biometric Ltd, UK) attached to the anterior aspect of the lower leg and foot. Ground contact of the heel during ground stepping was detected by a pressure-sensitive sensor (PH-463, DKH, Japan) placed under the heel. All signals from the right leg were sampled at 2 kHz using an A/D converter (WE 7000, Yokogawa Co. Ltd, Japan) and stored for later analyses.

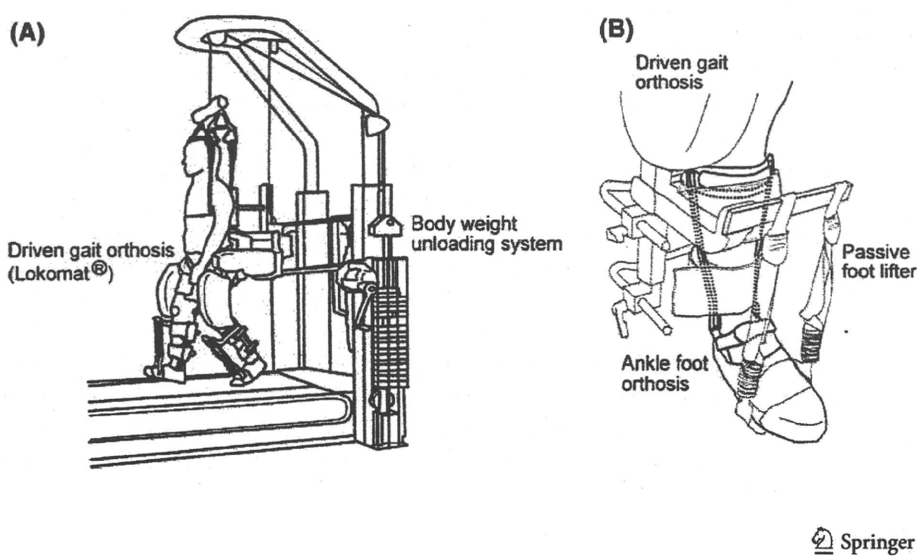


Fig. 1 **a** Schematic illustration of the experimental set-up for this study. Passive stepping was controlled by a driven gait orthosis (DGO; Lokomat®). Body weight of a subject was unloaded by a parachute harness. **b** Schematic illustration of the ankle joint with a passive foot lifter (spring-assisted elastic strap) and ankle foot orthosis

H-reflex recording

Before recording the Sol H-reflex, subjects had sufficient practice for passive stepping, while EMG signals of the lower limb muscles (rectus femoris, biceps femoris, medial gastrocnemius, and TA muscles) were continuously displayed on an oscilloscope (TDS 3014B, Tektronix, USA). During the H-reflex recording, an experimenter monitored the EMG activities in these muscles on a computer screen. When EMG activity during passive stepping was observed by the visual inspection, stimulation for the H-reflex was interrupted and the subject was instructed to relax. After disappearance of the EMG activity, stimulation was applied again.

The Sol H-reflex was elicited by stimulating the posterior tibial nerve (1-ms square pulse) using a cathode in the popliteal fossa and an anode placed over the patella with an electrical stimulator (SEN-7203, Nihon Kohden, Japan). During passive stepping, an output signal from the Lokomat® system was used as a trigger signal for electrical stimulation. The output signal was generated at a constant angle of right hip joint during stepping. Before the H-reflex recording, subjects performed passive ground stepping to determine six trigger delays for the application of stimulation after the output signal. The six trigger delays corresponded to six different step phases that were equivalent to the early-, mid-, and late-phases of the stance and swing. Because the hip and knee trajectories during passive stepping were under direct computer control for both stepping conditions, no differences were observed in the hip and knee joint angles at the determined stimulation timing between ground stepping and air stepping. By a stimulation method using a hip joint signal from the DGO, Querry et al. (2008) reported that stimulation accuracy was within 0.5° for a defined hip joint position. During passive stepping, the stimulation was randomly delivered at six predetermined phases of a step cycle with greater than 5-s intervals. The constancy of the M-wave size normalized to the maximum M-wave (Mmax) was needed for recording the H-reflexes evoked by the same stimulus strength during stepping. Since there is possibility that the amplitude of the Mmax itself vary considerably during walking (Simonsen and Dyhre-Poulsen 1999), measurements of the Mmax amplitudes by supramaximal stimulation in each phase for both loading conditions were performed. The H-reflexes were recorded at the stimulus intensity with the M-wave amplitude of ~10% Mmax in each phase. In addition to passive stepping, the amplitudes of the Mmax and H-reflex were measured at standing with 40 and 100% body weight unloading (ground standing and air standing, respectively). The stimuli were applied randomly with greater than 5-s intervals during the standing conditions. More than four sweeps for Mmax and more than ten sweeps for acceptable H-reflex in each measurement were recorded. For the SCI patients, the H-reflex was recorded only during ground stepping and standing.

As an additional experiment, recruitment curves of the M-wave and H-reflex were recorded in three of ten healthy subjects. Recording of the recruitment curve was performed at air standing and at the stance and swing phases of air stepping. The stimulus intensity was gradually increased from below the threshold of the H-reflex to supra-maximum stimulation of the M-wave. Three responses were recorded at each stimulus intensity.

Data analysis and statistics

The sizes of M-wave and H-reflex were assessed by peak-to-peak amplitudes, which were normalized to the respective Mmax amplitude recorded at each standing condition and at each phase of the stepping condition. The background EMG activities in the Sol and TA muscles were determined as the root mean square values of the EMG signals for 50 ms just prior to stimulation. If TA background activity was observed at the timing of H-reflex stimulation, the reflex response was removed from the data analysis. The data are shown as mean ± SD. The M-wave, H-reflex, and background EMG levels during passive stepping were analyzed by a two-way repeated measure ANOVA with factors of loading (ground and air stepping) and phase (6 phases in the step cycle). When the assumption of sphericity by Mauchly's test was violated, Greenhouse-Geisser adjustments were applied to adjust the degrees of freedom. When statistical significance was detected by ANOVA, post hoc multiple comparisons (Bonferroni) were used to identify the significant differences. The paired Student's t test was used for comparing between the two standing conditions. Statistical analyses were not performed for the data measured for the SCI patients. A statistical significant level was set at $P < 0.05$ in all cases.

Results

Figure 2 shows the EMG waveforms in the Sol and TA muscles, the angles at the hip, knee, and ankle joints, as well as the pressure-sensitive sensor signal at both loading conditions during passive stepping in a healthy subject. In this figure, 10 waveforms were superimposed based on the initiation of hip extension in the right limb. Duration of the one-step cycle was 2,750 ms at 1.5 km/h. Since the hip- and knee-joint trajectories of the DGO were computer-controlled, the joint movements were highly repeatable, and no difference was observed between the two loading conditions. The trajectory of the ankle joint showed a

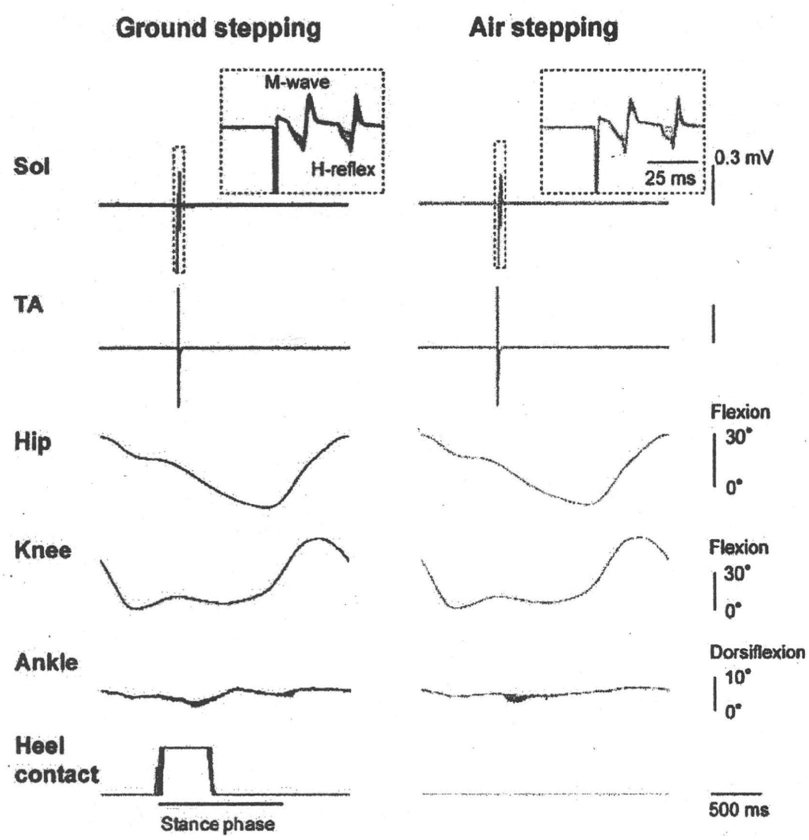


Fig. 2 Typical example of superimposed waveforms (10 sweeps) from the electromyographs (EMGs) of the soleus (Sol) and tibialis anterior (TA) muscles, angles of hip, knee, and ankle joints, and heel contact during ground and air stepping in a healthy subject. The electrical stimuli were applied at the early stance phase. The evoked responses are enlarged at each stepping condition

similar pattern in both loading conditions because of foot orthosis, but the angular variation in the stance phase increased slightly during ground stepping. No EMG activity was observed in the Sol and TA muscles at both loading conditions during passive stepping in this subject. Stimuli to evoke the H-reflex were applied at the early stance phase, and the evoked H-reflex waveforms are shown in the enlarged display. Due to the minimal variability in joint trajectories and muscle activities during stepping, the H-reflexes with a constant M-wave size could be easily evoked.

Figure 3 shows raw waveforms of the Mmax and H-reflex responses at two standing conditions and at six phases of two stepping conditions from a healthy subject. The stimulus intensity for the H-reflex was adjusted to evoke an M-wave size of 10% Mmax. Five sweeps for the Mmax and ten sweeps for the H-reflex are superimposed at each condition in this figure. There was phase-dependent modulation of the H-reflex excitability in both passive stepping conditions, showing that the Sol H-reflex at the early swing phase was markedly suppressed. In the present study, the H-reflex at the swing phase was completely suppressed in half of the subjects, while a small H-reflex response at the swing phase was observed in the remaining subjects.

Figure 4 represents the mean values of M-wave, H-reflex, and background EMG levels during passive stepping for all healthy subjects. The mean background EMG level of the Sol muscle, normalized to Mmax, was 0.09% Mmax for air standing and 0.08–0.09% Mmax through the six phases of air stepping. During ground standing, the background EMG level in the Sol was 0.0046 ± 0.0018 mV, which corresponding to 0.13% Mmax. Small muscle activation in the Sol was observed at the late-stance phase of ground stepping in four of ten subjects. The mean Sol background levels were 0.08–0.11% Mmax through the six phases of ground stepping. Although Sol background EMG levels tended to be larger during ground stepping than during air stepping, two-way ANOVA tests (2 loading conditions $\times$ 6 phases) showed that the main effect of loading was not significant ($F_{1,9} = 5.08$, $P = 0.051$). No significant effects of the phase and loading $\times$ phase interaction were observed in the Sol background EMG. Similarly, background EMG levels in the TA were not significantly different in loading conditions and among the step phases. In the present study, the Mmax size was not measured in the TA muscle. Instead, the EMG level during maximum voluntary contraction was recorded. The TA background levels, normalized by the EMG level during maximum voluntary

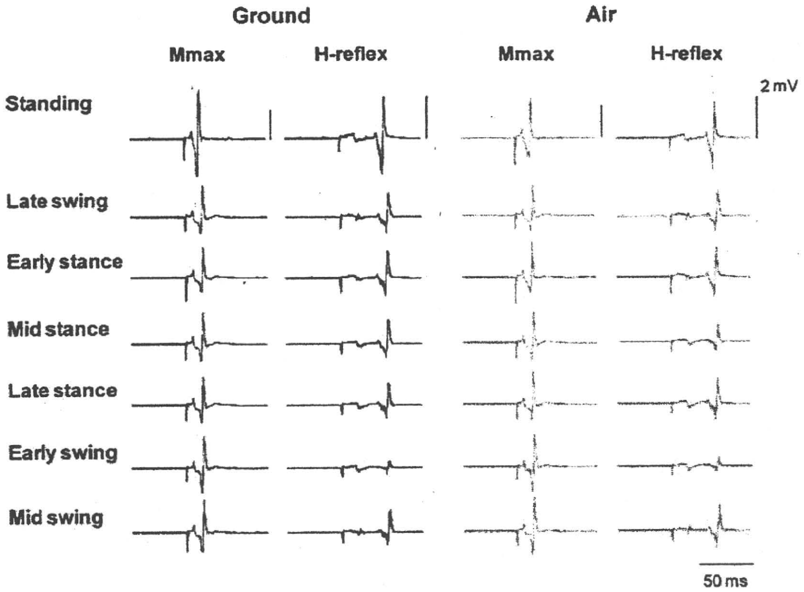


Fig. 3 Typical superimposed maximum M-wave (Mmax, 5 sweeps) and H-reflex (10 sweeps) waveforms with an M-wave size of 10% Mmax at standing and at six phases of passive stepping in a healthy subject

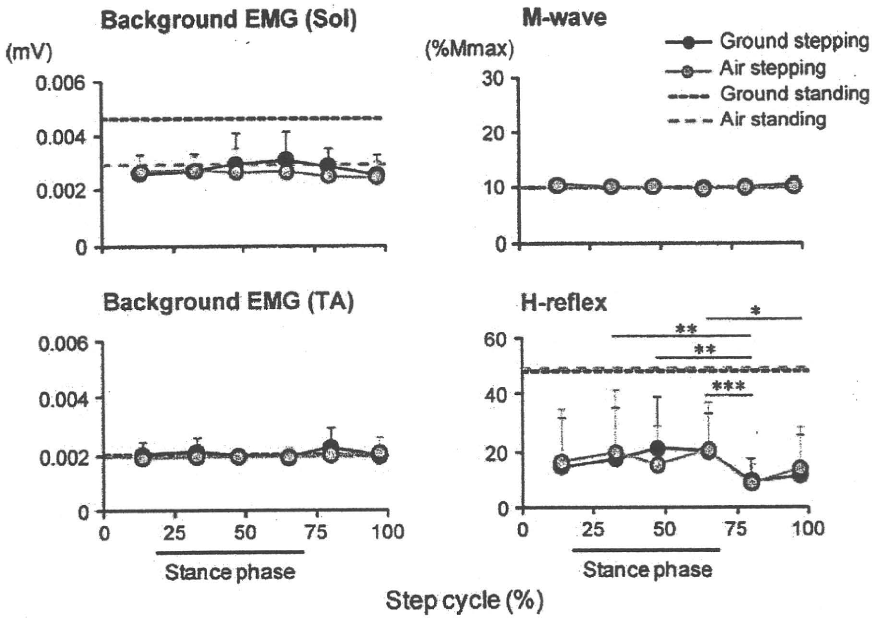


Fig. 4 Mean M-wave, H-reflex, and background EMG activity in the soleus (Sol) and tibialis anterior (TA) muscles at each phase during passive ground and air stepping in all healthy subjects. Significant difference between step phases during passive stepping, * $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$. The values for two unloading standing conditions are shown by *black* (ground standing) and *gray* (air standing) *dashed lines*

contraction, were less than 1% for both the standing conditions and at each phase for both stepping conditions.

In standing condition, the mean sizes of the H-reflex with M-wave corresponding to 10% of Mmax were not significantly different between ground and air standing (ground standing 48.3 ± 22.4% Mmax, air standing 49.1 ± 25.8% Mmax). Although the M-wave size was constant throughout the recording, the H-reflexes evoked at each phase during passive stepping were significantly inhibited compared to those at standing. Two-way ANOVA for the H-reflex during stepping revealed a significant main effect for phase ($F_{5,45} = 3.61$, $P < 0.05$), but not for loading ($F_{1,9} = 0.07$, $P > 0.05$), and no loading × phase interaction was observed ($F_{5,45} = 1.63$, $P > 0.05$). These results suggest that the H-reflex excitability was modulated in a phase-dependent manner during passive stepping, but no difference between ground and air stepping conditions. Post hoc test demonstrated that the Sol H-reflex was significantly inhibited at the early swing phase compared to the three stance phases ($P < 0.01$), and at the mid-swing phase compared to the late-stance phase ($P < 0.05$).

Figure 5 shows the M-wave and H-reflex (H-M) recruitment curves at air standing and at the stance and

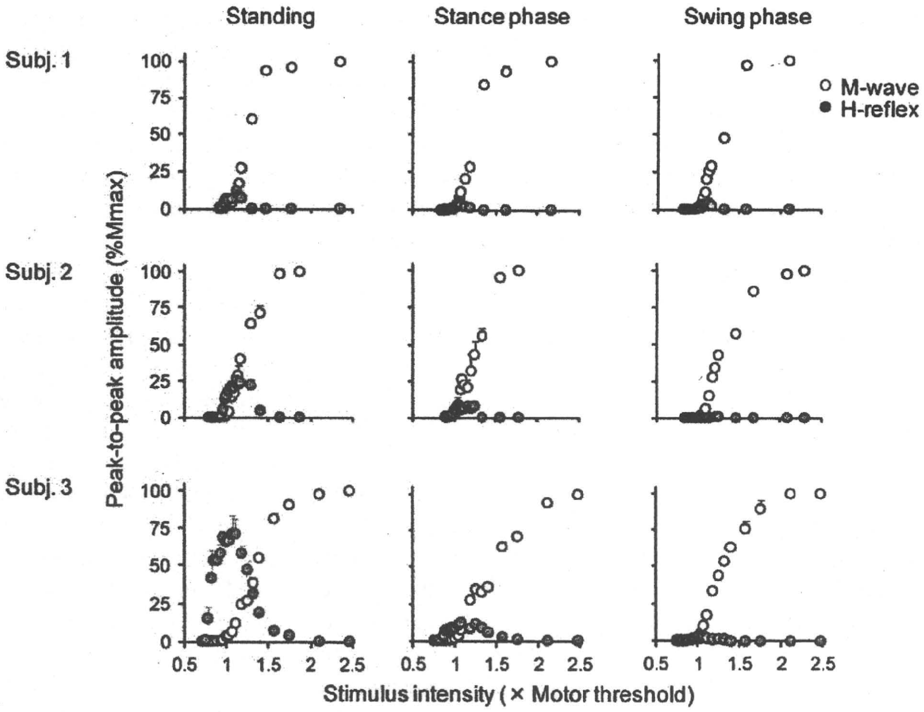


Fig. 5 Examples of H–M recruitment curves at standing and at the swing and stance phases of passive air stepping from three healthy subjects. Each plot shows the mean value of three responses at each stimulus intensity

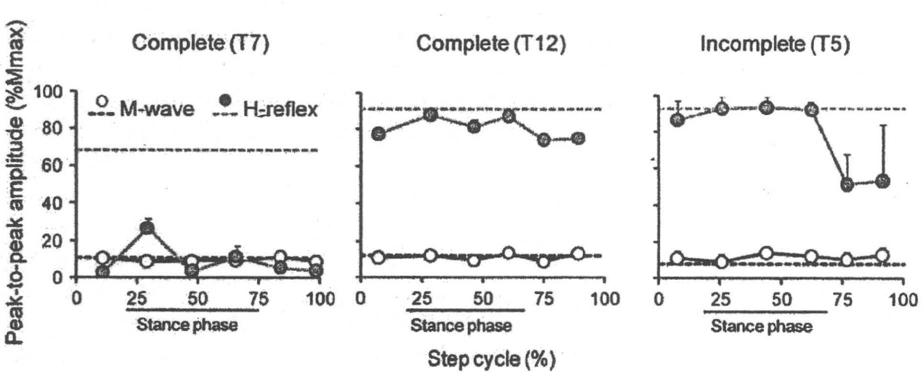


Fig. 6 Mean M-wave and H-reflex amplitudes during passive ground stepping in three spinal cord-injured (SCI) patients. The black and gray dashed lines represent the mean amplitudes of the M-wave and H-reflex at ground standing, respectively

swing phases of air stepping from three healthy subjects. The angle of the hip joint was similar at the instant when these H-M recruitment curves were measured. Although there were marked differences in the maximal H-reflex amplitude at the standing posture among the three subjects, the maximal H-reflex amplitude decreased from the standing to the stance and to the swing phase in all subjects. In addition, it was evident that the H-reflex recruitment curves at the swing phase of passive stepping were reduced across a wide range of stimulus intensities.

Furthermore, in the present study, we recorded H-reflex during ground stepping for two complete and one incomplete SCI patients. These patients showed elevated H-reflex amplitudes at ground standing (mean value in three SCI patients: 84.3 ± 12.0% Mmax, Fig. 6). During ground stepping, spastic muscle activities were rarely observed for a short time at the beginning of stepping and after the electrical stimulation in two SCI patients. Therefore, we applied the simulation when no spastic EMG activation was observed. Although the extent of H-reflex inhibition during ground stepping differed among the three patients (Fig. 6), the mean H-reflex amplitude at three (early-, mid-, and late-) swing phases was smaller than that at three stance phases (mean values at the stance and swing phases for all SCI patients: 64.2 ± 43.5 and 47.9 ± 38.3% Mmax, respectively).

Discussion

The aim of this study was to clarify the effect of peripheral sensory inputs on the Sol H-reflex excitability during walking. In the present study, passive stepping was conducted using a DGO in healthy subjects and SCI patients who were asked to keep their lower-limb muscles relaxed during stepping. During passive stepping, the H-reflexes were inhibited at the swing phase compared to the stance phase, which showed phase-dependent modulation. No significant difference was observed in the H-reflex excitability between different body weight unloading conditions for passive stepping.

Mechanism for H-reflex modulation during passive stepping

A constant M-wave size normalized to Mmax is typically used as an index of stimulus intensity to excite the group Ia afferents for H-reflex experiments to dynamic movement in humans. However, there is a possibility that the Mmax amplitude itself differs among the step phases due to changes in muscle geometry with respect to the surface electrodes and stimulus efficacy to nerves (Simonsen and Dyhre-Poulsen 1999). Therefore, in the present study, the Mmax size was recorded at each phase of stepping, and the H-reflex was evoked at a stimulus intensity in which the M-wave amplitude was 10% Mmax (Fig. 3). Compared to the stance phase, the Sol H-reflex with a constant M-wave was markedly reduced in the swing phase of the passive stepping (Fig. 4). Moreover, the H-reflex amplitudes across a wide range of stimulus intensities were inhibited from the standing posture to the stance phase to the swing phase (Fig. 5), indicating that any change in the H-reflex amplitude during passive stepping was not due to a change in the stimulus efficacy to the posterior tibial nerve.

The present study attempted to reduce the effect of voluntary command on the Sol H-reflex during walking by applying DGO to the healthy subjects. Indeed, no TA or Sol EMG activity was observed during ground and air stepping, except the Sol EMG activity at the latter-stance phase of ground stepping (Figs. 2, 4). Thus, with substantially reduced descending drive to the stepping task in healthy subjects, the results obtained during passive stepping would provide the effect of peripheral sensory inputs on the H-reflex excitability. Moreover, we investigated the H-reflex modulation during passive stepping in the SCI patients whose commands from the supraspinal center were completely or incompletely interrupted. Three SCI subjects who showed exaggerated H-reflex excitability at standing posture revealed that the Sol H-reflex was inhibited at the swing phase of passive stepping (Fig. 6). Recently, Phadke et al. (2007) investigated the effects of different walking environments (treadmill with body weight support and manual assistance by trainers as well as overground walking with an assistive device and brace) on the Sol H-reflex in two phases of mid-stance and mid-swing in incomplete SCI patients. They found that the H-reflexes during mid-swing phase were smaller than those during mid-stance phase for both walking environments in incomplete SCI patients. While they investigated the Sol H-reflex during active voluntary stepping for incomplete SCI, we also found a similar tendency for H-reflex modulation in six phases during passive stepping in SCI patients.

From previous studies, it is known that passive movements around the hip or knee joint significantly inhibit the Sol H-reflex. For example, reflex inhibition appears at the flexion phase of hip and knee joints, and peaks close to full flexion during passive pedaling (Cheng et al. 1995). Knikou et al. (2007) reported that the passive flexion of the hip significantly depressed the Sol H-reflex excitability when SCI patients were in the supine position. Moreover, afferent signals from the quadriceps muscles in decerebrate cats inhibit the Sol H-reflex during locomotion (Misiaszek and Pearson 1997). These results suggest that the inhibition of the H-reflex arises from movement-elicited sensory receptors discharging at the hip and knee joints. The findings of the present study also demonstrated that the H-reflex inhibition increased at the early- and mid-swing phases when the hip and/or knee joints were flexed. Therefore, the source underlying the phase-modulation of the H-reflex during passive stepping can be attributed to the sensory inputs, which likely arise from muscle spindle primary endings when the joints are flexed. The modulation pattern of the Sol H-reflex during passive stepping was similar to that observed during normal walking (Capaday and Stein 1987; Simonsen and Dyhre-Poulsen 1999; Ethier et al. 2003). Therefore, it is considered that the modulation pattern in the Sol H-reflex excitability during normal walking might be largely formed by peripheral sensory inputs associated with the lower-limb movements.

It is widely thought that phase-modulation of the Sol H-reflex during normal walking is associated with presynaptic inhibition of the Ia terminal (Capaday and Stein 1987; Crenna and Frigo 1987; Zehr 2002). Because no EMG activity in the TA muscle was observed during stepping in the present study, it appears that postsynaptic effect from reciprocal inhibition might play a minor role in the Sol H-reflex inhibition. In addition, it has been reported that the Sol H-reflex was still inhibited during passive movement with voluntary tonic contraction of the Sol muscle (Brooke et al. 1995; Misiaszek et al. 1995). When excitability in the motoneuron pool is stabilized by tonic contraction during passive movement, the postsynaptic effects on the H-reflex inhibition are likely to be minimized. Thus, it appears that the major source of the H-reflex modulation during passive stepping can be attributed to the presynaptic inhibition. However, the postsynaptic effect on the H-reflex modulation cannot be excluded in the present study. Because passive stepping was performed without the background EMG, any change in the subliminal fringe of the motoneuron during passive stepping was unknown. The change in the resting membrane potential of the motoneuron dependent on a step cycle might be partly involved in the H-reflex modulation during passive stepping.

Methodological limitations

In the present study, passive stepping was used to investigate the role of sensory inputs on the H-reflex modulation during walking. Although it appears that the modulation observed in the present study was generated by processes within the spinal cord, a few methodological limitations should be noted for the interpretation of these results. By instructing the subjects to relax during passive stepping, EMG activities in the lower limb muscles were not observed. However, the nearly complete disappearance of the EMG activity may not be a sufficient criterion to make the claim that there was no influence of descending drive on the H-reflex excitability. Also, because the sensory information that the subjects received during passive stepping with body weight unloading was different from that during normal walking, even a minor effect from the cortex due to the novel sensory perceptions during passive stepping may be related to modulations of the H-reflex. Furthermore, a possible implication of descending drive on the H-reflex excitability cannot be completely excluded, even in complete SCI patients, as it has been reported that a small percentage of individuals designated as complete SCI converted to incomplete SCI within 1 year after injury (Marino et al. 1999). Thus, the effect of the supraspinal input on the H-reflex modulation might not have been entirely eliminated in the present study.

Comparisons of reflex modulation during passive stepping and normal walking

In the present study, half of the subjects showed almost complete suppression at the swing phases of both stepping conditions, while the remaining subjects showed small H-reflex responses. Such inter-individual differences in the H-reflex inhibition have also been observed during normal walking (Simonsen et al. 2002). However, the amount of change in the H-reflex amplitude through the step cycle during passive stepping appeared to be less than that observed during normal walking. During the stance phase of normal walking, the Sol H-reflex increases progressively, nearly in parallel with the Sol background EMG (Capaday and Stein 1986). Ethier et al. (2003) demonstrated that, in most subjects, the H-reflex amplitude at late-stance during normal walking was larger than that during standing. However, the present study found that the H-reflex excitability at the stance phase in both loading conditions during passive stepping was significantly lower than that during standing. The lower H-reflex excitability in the stance phase during passive stepping when compared with normal walking might be attributed to little or no Sol EMG activity. As for the swing phase of normal walking, the H-reflex is almost completely suppressed due to the relevance of reciprocal inhibition, while TA is active (Ethier et al. 2003). Schneider et al. (2000) also observed that during the swing phase in a knee-locked walking task, the H-reflex inhibition was correlated with the TA EMG activity, and suggested that the modulation pattern during human walking follows the centrally produced reciprocal inhibition. In the present study, the H-reflexes were not completely suppressed at the swing phase of passive stepping in half of the subjects, presumably due to the absence of TA EMG activity. Thus, the EMG activity of the Sol and TA muscles that occurs normally during walking might quantitatively contribute to phase-modulation of the H-reflex. In addition, the difference in stepping speed between normal walking and this passive stepping condition might also be involved in the amount of the H-reflex inhibition during the swing phase because attenuation in the Sol H-reflex would depend on the velocity of the joint movement (McIlroy et al. 1992; Cheng et al. 1995).

Effect of load-related sensory inputs on reflex modulation during passive stepping

Inputs from load-related receptors during normal locomotion are well recognized to be significant for neural control (Harkema et al. 1997; Van de Crommert et al. 1998; Dietz and Duysens 2000; Dietz et al. 2002). Dietz et al. (2002) showed that afferent inputs from the hip joint in combination with those from the load receptors play crucial roles in generation of locomotor activity in SCI patients. However, the extent to which the Sol H-reflex is modulated by the load-related sensory inputs during walking is unclear. In the present study, there was no foot contact with the treadmill during air stepping (Fig. 2). Knikou and Conway (2001) observed that during sitting, the Sol H-reflex is inhibited by pressure (15–80 N) applied to the foot sole in healthy subjects and SCI patients. Therefore, we hypothesized that the Sol H-reflex during passive stepping would be inhibited by body weight loading. However, no significant difference was observed in the H-reflex between air and ground stepping in the present study (Fig. 4), i.e., no effect of the load-related sensory information was observed on the Sol H-reflex excitability during passive stepping. It is well known that the Sol H-reflex excitability is inhibited in a task-dependent manner from lying to sitting, to standing, and to walking (Capaday and Stein 1987; Crenna and Frigo 1987; Brooke et al. 1991; Zehr 2002; Stein and Thompson 2006). Similarly, the difference of the load-related effect on the H-reflex excitability between sitting and passive stepping could be explained by task dependency. Alternatively, the different loading effects on the H-reflex between sitting and stepping may be attributed to the differences in pressure to the foot sole. While tonic pressure was applied to the metatarsal region of the foot sole in the sitting condition (Knikou and Conway 2001), during walking, the focus of the pressure on the foot sole in stance phase moved from the heel toward the toe, and the amount of foot pressure was also changed through the stance phase (Nakajima et al. 2008). Thus, the difference in the sensory feedback provided by the pressure on the foot sole might possibly account for the inconsistent results for the loading effects on the H-reflex excitability between two different tasks.

Regarding the load effect on the transmission in other afferent pathways, Faist et al. (2006) reported that Ib inhibition from the stimulation of the gastrocnemius nerve to the Sol H-reflex was reduced by loading of the leg, regardless of motor tasks. In the present study, although the load-related afferent inputs were involved during ground stepping, there were no significant differences in modulation of the Sol H-reflex between ground and air stepping. However, it was observed that slight difference of the H-reflex amplitude between two stepping conditions was observed only in the mid-stance phase (Fig. 4). During passive stepping on a treadmill, foot sole pressure was the greatest in this phase (Nakajima et al. 2008). Regarding the effect of loading on Ib inhibition during stepping, additional investigations may be required.

As for the effect of load on the cutaneous reflex pathway, Bastiaanse et al. (2000) previously suggested that load-related afferent inputs were involved in the regulation of the cutaneous reflex in the lower limb muscles evoked by sural nerve stimulation during normal walking. In addition, using the same DGO as in the present study, we have recently observed strong facilitation of the cutaneous reflex in the TA muscle during the late-stance to early swing phase of passive ground stepping, but not during passive air stepping (Nakajima et al. 2008). These results suggest that the load-related sensory inputs play a key role in modulation of cutaneous reflexes during walking. In contrast, no significant difference was observed in the H-reflex between ground and air stepping in the present study. Also in a recent study by Knikou et al. (2009), it was reported that Sol H-reflex modulation remained constant across 0, 25, and 50% body weight support levels during treadmill walking for healthy subjects. Thus, it appears that the effects of load-related sensory inputs to the reflex responses during stepping are different between the cutaneous and the H-reflex pathways.

Acknowledgments This study was partially supported by the Grant-in-Aid for Young Scientists (B) (No. 19700460) from the Japanese Ministry of Education, Culture, Sports, Science and Technology (MEXT) and Global COE program on "Cybernics: fusion of human, machine, and information systems" from the Japan Society for the Promotion of Science.

References

Bastiaanse CM, Duysens J, Dietz V (2000) Modulation of cutaneous reflexes by load receptor input during human walking. Exp Brain Res 135:189–198

Brooke JD, Collins DF, Boucher S, McIlroy WE (1991) Modulation of human short latency reflexes between standing and walking. Brain Res 548:172–178

Brooke JD, Misiaszek JE, Cheng J (1993) Locomotor-like rotation of either hip or knee inhibits soleus H reflexes in humans. Somatosens Mot Res 10:357–364

Brooke JD, Cheng J, Misiaszek JE, Lafferty K (1995) Amplitude modulation of the soleus H reflex in the human during active and passive stepping movements. J Neurophysiol 73:102–111

Capaday C, Stein RB (1986) Amplitude modulation of the soleus H-reflex in the human during walking and standing. J Neurosci 6:1308–1313

Capaday C, Stein RB (1987) Difference in the amplitude of the human soleus H reflex during walking and running. J Physiol 392:513–522

Cheng J, Brooke JD, Misiaszek JE, Staines WR (1995) The relationship between the kinematics of passive movement, the stretch of extensor muscles of the leg and the change induced in the gain of the soleus H reflex in humans. Brain Res 672:89–96

Colombo G, Joerg M, Schreier R, Dietz V (2000) Treadmill training of paraplegic patients using a robotic orthosis. J Rehabil Res Dev 37:693–700

Crenna P, Frigo C (1987) Excitability of the soleus H-reflex arc during walking and stepping in man. Exp Brain Res 66:49–60

Dietz V, Duysens J (2000) Significance of load receptor input during locomotion: a review. Gait Posture 11:102–110

Dietz V, Müller R, Colombo G (2002) Locomotor activity in spinal man: significance of afferent input from joint and load receptors. Brain 125:2626–2634

Duysens J, Pearson KG (1980) Inhibition of flexor burst generation by loading extensor muscles in walking cats. Brain Res 187:321–332

Duysens J, Clarac F, Cruse H (2000) Load-regulating mechanisms in gait and posture: comparative aspects. Physiol Rev 80:83–133

Ethier C, Imbeault MA, Ung V, Capaday C (2003) On the soleus H-reflex modulation pattern during walking. Exp Brain Res 151:420–425

Faist M, Hoefer C, Hodapp M, Dietz V, Berger W, Duysens J (2006) In humans Ib facilitation depends on locomotion while suppression of Ib inhibition requires loading. Brain Res 1076:87–92

Garrett M, Kerr T, Caulfield B (1999) Phase-dependent inhibition of H-reflexes during walking in humans is independent of reduction in knee angular velocity. J Neurophysiol 82:747–753

Grey MJ, Nielsen JB, Mazzaro N, Sinkjær T (2007) Positive force feedback in human walking. J Physiol 581:99–105

Harkema SJ, Hurley SL, Patel UK, Requejo PS, Dobkin BH, Edgerton VR (1997) Human lumbosacral spinal cord interprets loading during stepping. J Neurophysiol 77:797–811

Knikou M, Conway BA (2001) Modulation of soleus H-reflex following ipsilateral mechanical loading of the sole of the foot in normal and complete spinal cord injured humans. Neurosci Lett 303:107–110

Knikou M, Schmit BD, Chaudhuri D, Kay E, Rymer WZ (2007) Soleus H-reflex excitability changes in response to sinusoidal hip stretches in the injured human spinal cord. Neurosci Lett 423:18–23

Knikou M, Angeli CA, Ferreira CK, Harkema SJ (2009) Soleus H-reflex modulation during body weight support treadmill walking in spinal cord intact and injured subjects. Exp Brain Res 193:397–407

Lavoie BA, Devanne H, Capaday C (1999) Differential control of reciprocal inhibition during walking versus postural and voluntary motor tasks in humans. J Neurophysiol 78:429–438

Marino RJ, Ditunno JF, Donovan WH, Maynard F (1999) Neurologic recovery after traumatic spinal cord injury: data from the model spinal cord injury systems. Arch Phys Med Rehabil 80:1391–1396

McIlroy WE, Collins DF, Brooke JD (1992) Movement features and H-reflex modulation. II. Passive rotation, movement velocity and single leg movement. Brain Res 582:85–93

Misiaszek JE, Pearson KG (1997) Stretch of quadriceps inhibits the soleus H reflex during locomotion in decerebrate cats. J Neurophysiol 78:2975–2984

Misiaszek JE, Brooke JD, Lafferty KB, Cheng J, Staines WR (1995) Long-lasting inhibition of the human soleus H reflex pathway after passive movement. Brain Res 677:69–81

Nakajima T, Kamibayashi K, Takahashi M, Komiyama T, Akai M, Nakazawa K (2008) Load-related modulation of cutaneous reflexes in the tibialis anterior muscle during passive walking in humans. Eur J Neurosci 27:1566–1576

Pearson KG, Collins DF (1993) Reversal of the influence of group Ib afferents from plantaris on activity in medial gastrocnemius muscle during locomotor activity. J Neurophysiol 70:1009–1017

Phadke CP, Wu SS, Thompson FJ, Berhman AL (2007) Comparison of soleus H-reflex modulation after incomplete spinal cord injury in 2 walking environments: treadmill with body weight support and overground. Arch Phys Med Rehabil 88:1606–1613

Querry RG, Pacheco F, Annaswamy T, Goetz L, Winchester PK, Tansey KE (2008) Synchronous stimulation and monitoring of soleus H reflex during robotic body weight-supported ambulation in subjects with spinal cord injury. J Rehabil Res Dev 45:175–186

Schneider C, Lavoie BA, Capaday C (2000) On the origin of the soleus H-reflex modulation pattern during human walking and its task-dependent differences. J Neurophysiol 83:2881–2890

Simonsen EB, Dyhre-Poulsen P (1999) Amplitude of the human soleus H reflex during walking and running. J Physiol 515:929–939

Simonsen EB, Dyhre-Poulsen P, Alkjær T, Aagaard P, Magnusson SP (2002) Interindividual differences in H reflex modulation during normal walking. Exp Brain Res 142:108–115

Stein RB, Thompson AK (2006) Muscle reflexes in motion: how, what, and why? Exerc Sport Sci Rev 34:145–153

Van de Crommert HW, Mulder T, Duysens J (1998) Neural control of locomotion: sensory control of the central pattern generator and its relation to treadmill training. Gait Posture 7:251–263

Zehr EP (2002) Considerations for use of the Hoffmann reflex in exercise studies. Eur J Appl Physiol 86:455–468

Human Movement Science 29 (2010) 556–566

Contents lists available at ScienceDirect

Human Movement Science

ELSEVIER

journal homepage: www.elsevier.com/locate/humov

Temporal correlations in center of body mass fluctuations during standing and walking

Masaki O. Abe [a,*], Kei Masani [b,c], Daichi Nozaki [d], Masami Akai [e], Kimitaka Nakazawa [f]

[a] *Action Lab, Department of Biology, Northeastern University, 134 Mugar Life Science Building, 360 Huntington Avenue, Boston, MA 02115, United States*
[b] *Rehabilitation Engineering Laboratory, Institute of Biomaterials and Biomedical Engineering, University of Toronto, 164 College Street, Toronto, ON, Canada M5S 3G9*
[c] *Rehabilitation Engineering Laboratory, Lyndhurst Centre, Toronto Rehabilitation Institute, 520 Sutherland Drive, Toronto, ON, Canada M4G 3V9*
[d] *Physical Education Laboratory, Graduate School of Education, University of Tokyo, 7-3-1 Hongo, Bunkyo-ku, Tokyo 113-0033, Japan*
[e] *Hospital of National Rehabilitation Center for Persons with Disabilities, 4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan*
[f] *Department of Life Sciences, Graduate School of Arts and Sciences, University of Tokyo, 3-8-1 Komaba, Meguro-ku, Tokyo 153-8902, Japan*

ARTICLE INFO

Article history:
Available online 6 July 2010

PsycINFO classification:
2330

Keywords:
Balance
Walking
Standing
Center of body mass
Temporal correlation

ABSTRACT

Body fluctuations during both quiet standing and walking exhibit temporal correlations that reflect mechanisms of balance control. However, knowledge about the relationship between the temporal structures observed during standing and walking is limited. The goal of the present study was (1) to investigate temporal correlations in the fluctuations of the center of body mass acceleration (ACC) in standing and walking, and (2) to test the hypothesis that the degree of the temporal correlation for the two tasks is similar and correlated across participants. Seventeen young, healthy participants stood and walked for 10 min on a treadmill equipped with two force platforms. The temporal correlations of the ACC in the anteroposterior (ACC_{AP}), mediolateral (ACC_{ML}), and two-dimensional (ACC_{2D}) directions were evaluated using the scaling index (α) as calculated with Detrended Fluctuation Analysis. The scaling indices of ACC fluctuations during standing and walking were categorized as stationary signals which are temporally correlated ($0.5 < \alpha < 1.0$). Further, there were significant, positive correlations for ACC_{AP} and ACC_{2D} between the scaling indices during standing and walking. The results suggest that there are common characteristics in the balance control system for standing and walking,

* Corresponding author. Tel.: +1 617 373 5093; fax: +1 617 373 3724.
E-mail address: m.abe@neu.edu (M.O. Abe).

doi:10.1016/j.humov.2010.03.002