表4　MS の有無による各パラメータ値の比較

	男性 MS(−)	男性 MS(+)	p	女性 MS(−)	女性 MS(+)	p
症例数	160	35		261	10	
年齢(歳)	48.1 ± 11.8	50.8 ± 10.9	0.26	49.0 ± 10.6	54.2 ± 13.7	0.14
身長(cm)	169.4 ± 6.2	169.2 ± 4.6	0.84	156.9 ± 5.3	156.2 ± 5.0	0.68
体重(kg)	68.5 ± 9.2	77.2 ± 8.9	<0.0001	55.1 ± 8.6	68.5 ± 11.5	<0.0001
BMI(kg/m²)	23.9 ± 3.0	26.9 ± 2.9	<0.0001	22.4 ± 3.3	28.0 ± 3.2	<0.0001
腹囲(cm)	83.5 ± 8.1	93.2 ± 6.6	<0.0001	76.4 ± 9.1	97.0 ± 3.8	<0.0001
ヒップ囲(cm)	93.4 ± 5.4	98.2 ± 5.4	<0.0001	91.7 ± 6.2	100.4 ± 6.9	<0.0001
体脂肪率(96)	19.2 ± 4.4	22.4 ± 6.0	0.0004	24.1 ± 6.4	33.1 ± 5.0	<0.0001
baPWV(cm/s)	1354.5 ± 222.6	1486.7 ± 208.1	0.0015	1256.2 ± 187.6	1648.0 ± 447.1	<0.0001
ABI	1.14 ± 0.08	1.13 ± 0.08	0.64	1.11 ± 0.07	1.12 ± 0.08	0.71
収縮期血圧(mmHg)	125.3 ± 13.0	138.7 ± 15.0	<0.0001	117.2 ± 14.4	148.6 ± 16.5	<0.0001
拡張期血圧(mmHg)	76.1 ± 9.0	85.8 ± 9.4	<0.0001	69.2 ± 9.9	88.4 ± 6.6	<0.0001
中性脂肪(mg/dl)	108.3 ± 61.4	241.4 ± 139.0	<0.0001	84.0 ± 45.1	158.5 ± 63.4	<0.0001
HDL-cho(mg/dl)	61.7 ± 15.5	52.6 ± 27.5	0.009	73.2 ± 17.7	53.5 ± 11.2	0.0005
血糖(mg/dl)	96.1 ± 11.3	105.9 ± 18.5	<0.0001	90.7 ± 7.9	105.8 ± 16.7	<0.0001

平均値±標準偏差

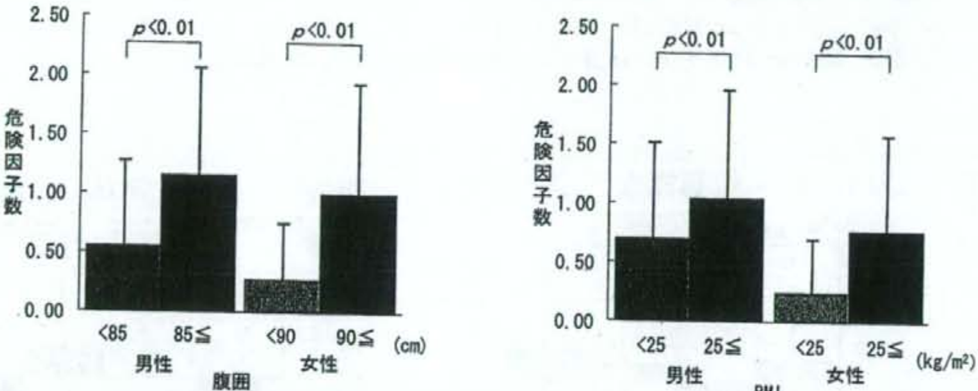

図1　肥満指標からみた危険因子数

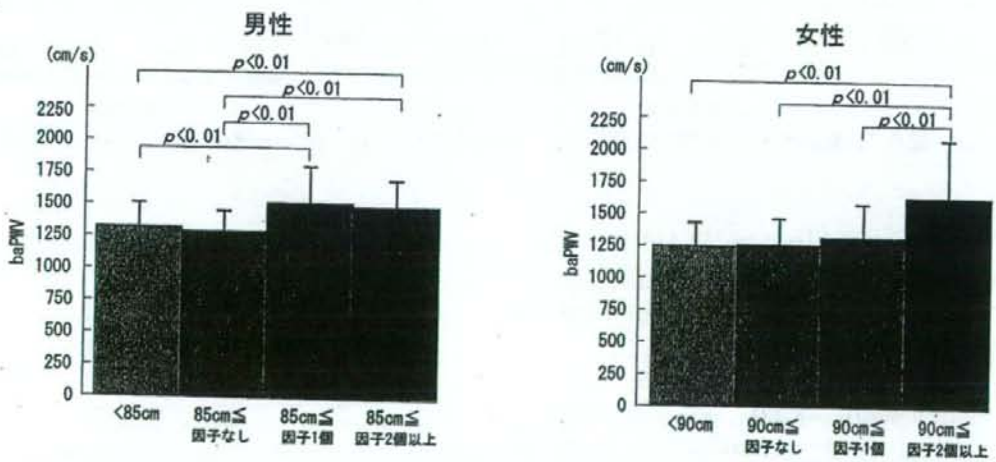

図2　腹囲別，メタボリックシンドローム構成因子数と baPWV

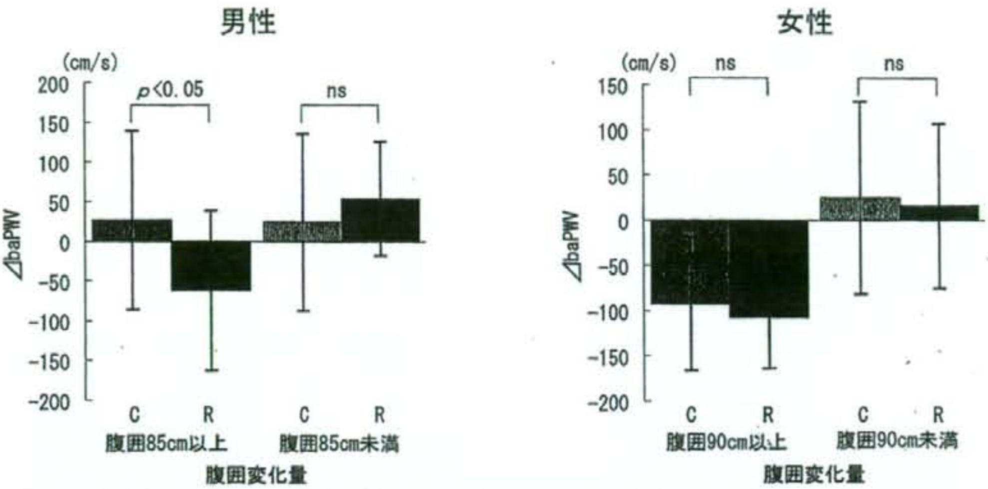

図3　腹囲変化量と baPWV 変化量（R：腹囲変化量≦－3cm，C：腹囲変化量＞－3cm）

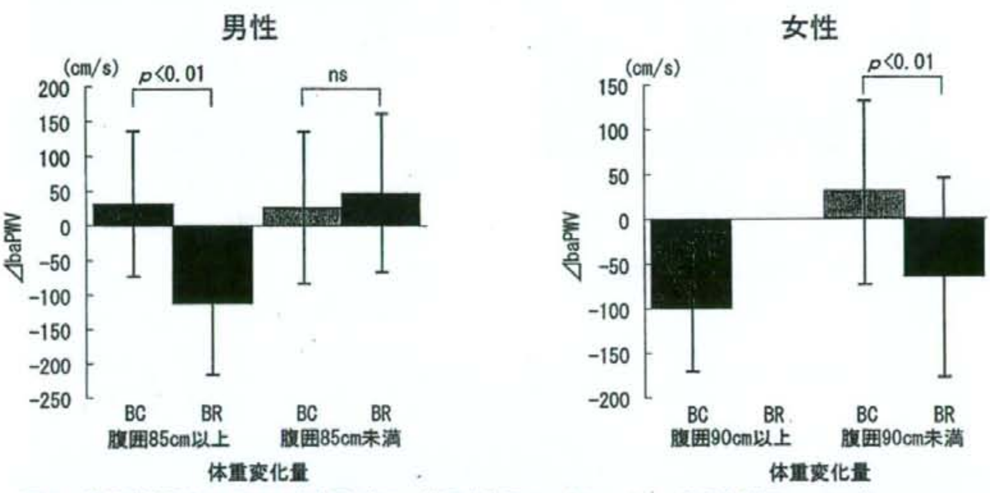

図4　体重変化量と baPWV 変化量（BR：体重変化量≦－3kg，BC：体重変化量＞－3kg）

1486.7 ± 208.1cm/s vs 非 MS 群：1354.5 ± 222.6cm/s；$p < 0.005$，女性 MS 群：1648.0 ± 447.1cm/s vs 非 MS 群：1256.2 ± 187.6cm/s；$p < 0.0001$）（表4）。

また，腹囲 85（90）cm 未満，85（90）cm 以上で高血圧，高血糖，脂質異常の危険因子保有数別の各群で baPWV を比較，検討すると，男性では，85cm 未満群，85cm 以上危険因子 0 群に比較して，85cm 以上危険因子 1 群，85cm 以上危険因子 2 以上群が有意に高値であった。女性では，90cm 未満群，90cm 以上危険因子 0 群，90cm 以上危険因子 1 群に比較して，90cm 以上危険因子 2 以上群が有意に高値であった（図2）。

対象 2 において，腹囲変化量と baPWV 変化量（⊿ baPWV）との関係について検討したところ，男性では，腹囲が 3 cm 以上減った群（R 群）と 3 cm 減らなかった群（C 群）に有意差を認めなかった。しかしながら，初回測定時腹囲 85cm 以上だった者で検討すると，R 群に比較して C 群の ⊿ baPWV が有意に高値であった（R 群：－ 62.0 ± 100.7cm/s vs C 群：26.5

± 112.7cm/s；$p < 0.05$)。女性では，R 群と C 群に有意差を認めず，初回測定時腹囲 90cm 以上だった者でも有意差は認められなかった（図 3）。

次に，体重変化量と⊿ baPWV との関係について検討したところ，男性では，体重が 3kg 以上減った群（BR 群）に比較して 3kg 減らなかった群（BC 群）の⊿ baPWV が有意に高値であった(BR:－55.3 ± 130.0 cm/s vs BC 群: 28.1 ± 107.0cm/s；$p < 0.01$)。初回測定時腹囲 85cm 以上だった者でも，BR 群に比較して BC 群の⊿ baPWV が有意に高値であった（BR 群:－112.2 ± 104.0cm/s vs BC:31.3 ± 104.6cm/s；$p < 0.0005$)。女性では，BR 群に比較して BC 群の⊿ baPWV が有意に高値であった（BR 群:－64.8 ± 106.9cm/s vs　BC 群：24.9 ± 104.3 cm/s；$p < 0.005$)（図 4）。初回測定時腹囲 90cm 以上は，BR 群が 1 名しかおらず検討できなかった。

考　　察

現在わが国では，メタボリックシンドロームが，腹部肥満を基盤に高血圧，高血糖，脂質異常などの危険因子が重なり合い，動脈硬化の形成に関与しているとして注目されている。以前，当センターにおけるメタボリックシンドロームの頻度を検討したが，男性 30.7%，女性 3.6%であった [4]。

フォルムを用いた baPWV は，動脈硬化の程度を反映し，簡便かつ非侵襲的に測定できることから全国に普及している。今回の横断的な検討において，MS 群は非 MS 群に比較して，有意に baPWV の高値が認められ，さらに，MS の危険因子数が多いほど高値を示した。健診受診者 185 名を対象として MS と baPWV を比較した佐々木らの研究 [5] においても，同様に，MS 診断の構成因子である肥満，高血圧，高血糖，脂質異常の累積数の増加により baPWV が亢進すると報告している。また，Tsubakimoto らも 525 名の日本人男女を対象に検討し同様な報告を行なっている [6]。つまり，MS の危険因子数が多いほど動脈硬化が進行していることが示唆された。

今回の検討で特筆すべき点は，縦断調査により腹囲，体重の変化量と baPWV の関係を明らかにできた点である。日本肥満学会（http://wwwsoc.nii.ac.jp/jasso/，accessed on July 13, 2007）では神戸宣言 2006 として，メタボリックシンドロームの予防，改善のために，食生活の改善と運動の増加を図り，まず 3kg の減量と 3cm の腹囲の減少を実現するサンサン運動を提案している。以前の私たちの調査から，少なくとも腹囲 3cm の減少により肥満男性では MS の改善を認めたことを報告した [7]。今回の縦断調査から，男性で腹囲 85cm 以上の者では少なくとも 3kg の減量と 3cm の腹囲の減少により baPWV が有意に減少したことを認めた。女性では，腹囲 90cm 以上の割合が少なかったため，男性のような関係が認められなかったものと思われ，今後，さらなる検討が必要と思われた。

baPWV は，MS の動脈硬化診断にも有用で，MS の危険因子数にも鋭敏に反応し，今後の MS の予防，改善の指標のひとつとして腹部肥満男性では，ますます利用価値が高まると思われた。

ま と め

岡山県南部健康づくりセンター利用者を対象として，MS と baPWV を用いた動脈硬化との関連を検討した結果，MS 群では非 MS 群に比較して動脈硬化が進行し，さらに危険因子の保有数が多いほど動脈硬化が進行していることが明らかとなった。また，男性の腹部肥満者では，日本肥満学会の神戸宣言 2006 の腹囲 3cm，体重 3kg の減少により，動脈硬化の改善が認められた。

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[RESEARCH REPORT]

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The Relationship Between Passive Ankle Plantar Flexion Joint Torque and Gastrocnemius Muscle and Achilles Tendon Stiffness: Implications for Flexibility

Flexibility is a component of fitness that is thought to be associated with athletic performance[29,34,36] and the incidence of muscular injury,[29,35] although research findings are equivocal in these aspects.[6] Flexibility is commonly evaluated by measuring maximal range of joint motion[2,20] or, alternatively, from the joint angle-passive torque relationship.[18] Flexibility evaluated using these 2 methods results in related values,[20] because the passive torque gradually increases as the joint is passively moved, until no further angular displacement is possible at the extreme position of joint motion.

• **STUDY DESIGN:** Experimental laboratory study.

• **OBJECTIVES:** We tested the hypothesis that the muscle fibers and the connective tendinous structures, combined in series, provide the resistance to passive joint movement at the ankle. We also determined the relative association between passive joint torque and each of these 2 elements.

• **BACKGROUND:** The reason for individual variation in joint flexibility or tightness is not clearly understood, but the influence of musculotendinous stiffness has been inferred.

• **METHODS AND MEASURES:** Each of the subjects (6 women and 6 men) was seated with the right knee extended and right ankle positioned at a 30°, 20°, 10°, 0°, −10°, −20°, and −30° (0, neutral position, positive values reflecting plantar flexion) angle while passive plantar flexion torque was measured. The distal muscle-tendon junction of the medial gastrocnemius was visualized by ultrasonography, and its positional change was defined as muscle belly length change. The whole muscle-tendon unit length change was estimated from joint angle changes, from which Achilles tendon length change was estimated.

• **RESULTS:** Both the muscle belly and tendon were significantly elongated as the ankle was dorsiflexed (at 0° the mean ± SD muscle belly elongation was 10.3% ± 1.8%, and the tendon elongation was 2.8% ± 1.2%, of the initial length of their respective structures measured at 30° of ankle plantar flexion), from which stiffness indices were determined both for muscle belly and tendon. The passive torque at 0°, −10°, −20°, and −30° was significantly correlated with the stiffness indices of the Achilles tendon (at 0°, $r^2 = 0.70$ and 0.62 for overall and specific stiffness, respectively; $P<.05$). A tendon stiffness index, separately obtained from tendon lengthening during maximal isometric contraction, was also correlated with passive ankle plantar flexion torque at 0°, −10°, −20°, and −30° (at 0°, $r^2 = 0.76$, $P<.05$). The specific stiffness index of the muscle belly was correlated ($r^2 = 0.47$, $P<.05$) with the passive ankle plantar flexion torque at 0°, but its overall stiffness index was not ($r^2 = 0.32$, $P>.05$).

• **CONCLUSION:** Results suggest that extensibility of the muscle-tendon unit of the Achilles tendon for the most part is related to passive ankle plantar flexion joint torque. *J Orthop Sports Phys Ther 2008;38(5):269-276. doi:10.2519/jospt.2008.2632*

• **KEY WORDS:** *dorsiflexion, flexibility, plantar flexors, stretching*

It has been suggested that joint flexibility or passive joint torque is influenced by musculotendinous structures around the joint.[2,4,20,30,31,34] The tendinous tissues are known to possess elastic characteristics[1,17,24] (ie, they are elongated by external forces and return to their initial length when the force is removed). Researchers have shown in vivo that the human Achilles tendon is stretched by 5% of its initial length during maximal isometric plantar flexion.[22] It has also been shown in animals[32] and also in humans[10,33] that both the tendinous tissues and muscle fibers are significantly elongated by passive stretching of the muscle-tendon unit. These findings suggest that the extensibility of muscle fibers and tendinous tissues is related to the passive joint torque. In the present study, we determined length changes of the gastrocnemius muscle belly and Achilles tendon during passive ankle joint angular displacement in dorsiflexion in humans. We tested a hypothesis that the muscle fibers and the connective tendinous structures combined in series provide the resistance to passive joint movement, and determined the relative association between passive

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joint plantar flexion torque and each of these 2 elements.

METHODS

Subjects

Six women and 6 men (age range, 19-38 years; mean ± SD height, 164.7 ± 10.2 cm; body mass, 62.4 ± 16.0 kg) volunteered and gave written informed consent to participate in the study. They had no apparent neurological, orthopedic, or neuromuscular problems. The study was approved by the Ethics Committee of the Department of Life Sciences, University of Tokyo. There was no statistical difference in age between women and men. Due to the limited number of subjects, all were pooled into 1 group.

Measurement of Passive Joint Torque To measure passive plantar flexion torques at different ankle joint positions, a specially designed apparatus was used in which the subject was seated with the right knee extended and the right ankle secured to a foot plate that was rotated and fastened at 10° increments (**FIGURE 1**). The foot plate was fixed with the ankle positioned at 30°, 20°, 10°, 0°, −10°, −20°, and −30° in this order. Zero degrees represents the neutral ankle position, positive and negative values are plantar flexion and dorsiflexion angles, respectively. At least 1 minute of rest, with the ankle returned to the 30° plantar flexion angle, was provided before testing at the next position. One minute of rest was adopted to ensure restoration of original musculotendon viscoelastic properties before testing at the next ankle position.[20] But excessive deformation of the tissues at higher load (further dorsiflexion) could alter the viscoelastic properties to the extent that they would not fully recover within 1 minute, and there is also a possibility of tendon creep at higher loads.[6,37] Therefore, we did not randomize the testing angle order in an attempt to minimize the carry-over effects by these factors. The subject was instructed to relax the muscles at each position before the measurement

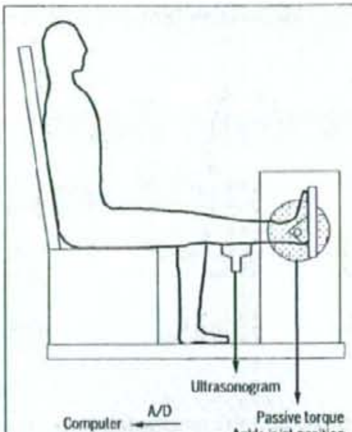

FIGURE 1. Experimental setup.

of passive plantar flexion torque. The room temperature was kept constant to minimize temperature-induced changes in elasticity of passive tissues.[30] The position at 30° plantar flexion was selected for the initial length of the muscle belly and the tendon because this position is where passive torque around the ankle has been shown to be near zero.[12,33,37] Our pilot work[23] showed that muscle activity (electromyographic response) was negligible up to the point at which the subjects started to feel a strong stretch, and that muscle activity increased when they started to have pain sensation. Therefore, care was taken in this study to avoid production of pain at each joint angle, and on the occasions that the subject felt a painful sensation, further testing was stopped. As a result, 2 subjects were not tested at −10°, −20°, and −30°, and 1 additional subject at −20° and another subject at −30°. For testing, each joint position was kept for 30 seconds and the passive torque measured at the end of that period was used for analysis.

Determination of Muscle and Tendon Elongation

The distal muscle-tendon junction (MTJ), defined in this study as the distal edge of the most distal fascicle of the medial gastrocnemius (MG) muscle, was visualized

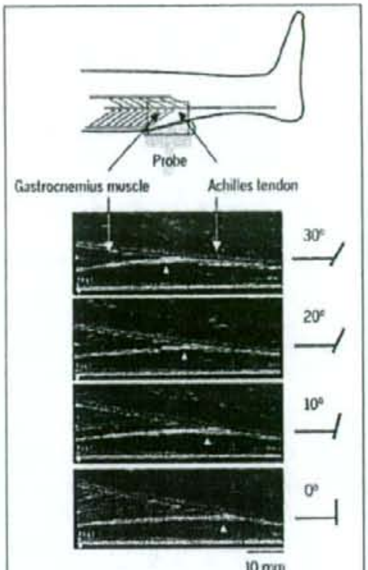

FIGURE 2. A schematic model of the calf (top) and ultrasonic images of the medial gastrocnemius muscle, with the distal muscle-tendon junction (white triangle) at ankle positions 30°, 20°, 10°, 0° (from top to bottom).

by an ultrasonic apparatus (SSD-900; Aloka, Tokyo, Japan) with an electronic linear array probe (7.5 MHz). The precision and linearity of ultrasonogram have previously been confirmed.[11,12] The probe was attached firmly with a strap over the MTJ so that the probe did not slide over the skin. A rubber water bag (MP-2463; Aloka, Tokyo, Japan) was placed between the probe and the skin to retain their close adherence and to improve visualization of the MTJ (**FIGURE 2**). The excursion of the MTJ for a given joint angular displacement (**FIGURE 2**) was determined to the nearest 1 mm as the lengthening of the MG muscle belly (including fascicles, proximal and distal aponeurosis, and proximal tendon). The accuracy and repeatability of this measurement has been confirmed in previous studies.[22,23]

Changes in the whole MG muscle-tendon unit (MTU) length were estimated from ankle joint angle changes. This was done based on a previously reported equation between relative MTU length

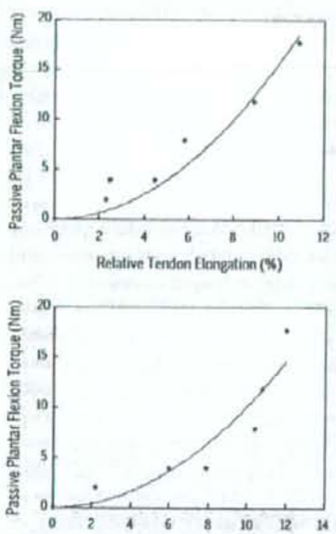

FIGURE 3. The relationship between passive torque and estimated tendon elongation and muscle belly elongation (example on 1 subject). Each relationship was fitted with a second-order polynomial equation, the coefficient of which was defined as the tendon stiffness index (TSIpas) and muscle belly stiffness index (MSIpas).

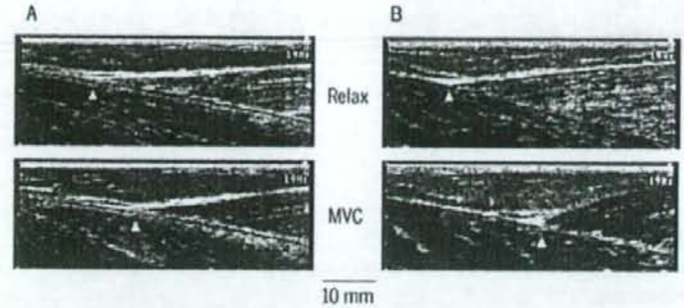

FIGURE 4. Ultrasonic images of the distal muscle-tendon junction at rest (top) and during maximal voluntary isometric contraction (MVC). The distal muscle-tendon junction is indicated by a white triangle in each image. Examples from 2 subjects with stiff (a) and compliant (b) tendon.

and knee and ankle joint angles, developed on cadavers.[8] The leg length, measured as the distance between popliteal crease and center of the lateral malleolus, was incorporated into the equation to obtain absolute MTU length changes for each subject. The tendon length change was then estimated by subtracting muscle belly length change from MTU length change. The tendon length defined here is the Achilles tendon length (the length of the distal outer tendon of the MG).[7] Excluded is the length of the distal aponeuroses, onto which the MG muscle fibers attach, which were included with the muscle belly. The MTJ of the lateral gastrocnemius is positioned almost at the same proximal-distal position as that of the MG (inspected by ultrasonogram and also from dissected cadaver specimens).[13] The aponeuroses of the medial and lateral gastrocnemii are laterally connected[7] and the 2 heads share the same Achilles tendon. These 2 heads also share some of the fascicles.[28] Therefore, we consider the measurements made in this study to be representative of the entire gastrocnemius muscle.

Evaluation of Tendon and Muscle Belly Stiffness

After testing, the MTJ, with the ankle positioned at 30°, was confirmed from ultrasonogram and marked with a pen on the skin, and the distance between MTJ and the calcaneal tuberosity (also confirmed by ultrasonogram) was measured over the skin as the initial tendon length. The relationship between length changes in tendon, normalized to the initial tendon length, and passive plantar flexion torque was obtained for each subject. This relationship was fitted with a second-order polynomial equation ($y = a \times x^2$) using a least-square regression (based on a previous study),[4,19] and the coefficient of equation was defined as a tendon stiffness index (TSIpas) (**FIGURE 3**). The selection of the equation is based on the curvilinear nature of the length-force properties of soft tissues at the low force portion ("toe region"),[17,20] and for the present study, both parameters were zero for the initial ankle position (30°). Also, the first derivative of the equation at the point of tendon elongation corresponding to 0° was determined (TSIpas-neutral). In addition, muscle belly length change was normalized to the initial muscle belly length, which was determined by subtracting tendon initial length from leg length. The normalized muscle belly length change was also related with passive torque to obtain muscle belly stiffness index (MSIpas) (**FIGURE 3**), using the same procedure as that of TSIpas. The first derivative of the equation for the ankle position at 0° was determined in a similar way (MSIpas-neutral). The reason for the selection of 0° for the tendon and muscle belly stiffness indices was that this was the furthest position from 30° at which all subjects' data were available.

After the above testing, the subject performed maximal voluntary isometric plantar flexion with the ankle secured at 0°. During contraction, MTJ shifted proximally (**FIGURE 4**) due to lengthening of the Achilles tendon.[22] This shift in MTJ was measured as the tendon elongation. The difference between the maximal isometric torque and passive torque at 0° was divided by the tendon elongation (normalized by the initial tendon length), which was defined as the tendon stiffness index by active contraction (TSIact). This calculation is based on the linear nature of the tendon force-length curve at high force portion ("linear region").[17]

Statistical Analyses

Values are expressed as means ± SD unless otherwise stated. Changes in torque

TABLE TISSUE LENGTHENING AND STIFFNESS INDICES OF MUSCLE BELLY AND TENDON (N = 12)*

	Muscle Belly	Tendon
Tissue lengthening from the initial position (30°)		
20°, mm (%)	7 ± 2 (4.1 ± 1.0)	1 ± 2 (0.6 ± 0.7)
10°, mm (%)	13 ± 2 (7.4 ± 1.0)	3 ± 2 (1.6 ± 0.9)
0°, mm (%)	18 ± 3 (10.3 ± 1.8)	6 ± 2 (2.8 ± 1.2)
−10° (n = 10), mm (%)	21 ± 4 (12.5 ± 2.6)	8 ± 3 (4.5 ± 1.6)
−20° (n = 9), mm (%)	24 ± 5 (14.1 ± 3.5)	12 ± 3 (6.5 ± 2.1)
−30° (n = 8), mm (%)	25 ± 5 (14.8 ± 3.6)	17 ± 4 (9.0 ± 2.4)
Tissue stiffness index		
Passive, overall (AU)†	0.14 ± 0.07	0.67 ± 0.54
Passive, specific (AU)‡	2.8 ± 1.2	3.4 ± 2.5
Active (AU)	...	18.1 ± 8.3

Abbreviation: AU, arbitrary unit. (The unit of tendon stiffness index is arbitrary, depending on parameters. For example, for passive overall stiffness it is Nm × %². See text for calculation of each parameter).

* *Values are mean ± SD. Positive ankle joint angle values are for plantar flexion and negative values for dorsiflexion.*

† *Passive overall: coefficient of regression between relative tissue lengthening and torque.*

‡ *Passive specific: first derivative of the regression curve at the point corresponding to ankle at 0°.*

and elongation of muscle belly and tendon at 20°, 10°, 0°, −10° (n = 10), −20° (n = 9), and −30° (n = 8), as compared to 30°, were tested for statistical significance by a 1-way repeated-measures ANOVA. For those variables for which a significant angle effect was found, Fisher's PLSD was used for a post hoc test. A linear regression analysis was performed for the relationships between the relative lengthening of muscle belly and tendon, TSIpas, TSIact, MSIpas, and the passive torque with the ankle at 0°, −10°, −20°, and −30°. In all cases, the level of significance was set at $P<.05$.

RESULTS

PASSIVE ANKLE PLANTAR FLEXION torque showed considerable interindividual variability (4-16 Nm at 0° and 18-61 Nm at −30°), but in all subjects it increased significantly as the ankle was passively dorsiflexed ($P<.05$ for all neighboring angles). The muscle belly was significantly lengthened as the ankle was passively dorsiflexed ($P<.05$ for all neighboring angles) (**TABLE**), with the magnitude of lengthening ranging from 12 to 21 mm (ankle positioned at 0°) and from 17 to 31 mm (at −30°), when compared to the initial ankle angle position (30°). The tendon was also lengthened ($P<.05$ for all neighboring angles) (**TABLE**), with the lengthening ranging from 2 to 10 mm at 0° and 12 to 23 mm at −30°, when compared to the initial ankle angle position of 30° plantar flexion. The averages ± SD lengthening of the muscle belly and tendon was 10.3% ± 1.8% and 2.8% ± 1.2%, respectively, at 0°, and these were negatively correlated with each other ($r^2 = -0.86$, $P<.05$).

During the maximal isometric contraction, the tendon was significantly lengthened by a mean ± SD of 17 ± 4 mm (9.0% ± 2.4% of the initial tendon length). Again, there was considerable intersubject variability in the magnitude of tendon lengthening. Typical ultrasound images are shown in **FIGURE 4**.

The TSIpas and TSIpas-neutral values were positively correlated ($r^2 = 0.64$, $P<.01$), and each measurement was significantly correlated with TSIact ($r^2 = 0.49$ and $r^2 = 0.34$, respectively; $P<.05$). All indices were positively correlated with the passive ankle plantar flexion torque measured at 0° ($r^2 = 0.70$, 0.62, and 0.76 for TSIpas, TSIpas-neutral, and TSIact, respectively; $P<.01$). Although for smaller numbers of subjects, TSIpas-neutral, and TSIact values were also significantly correlated with the passive ankle plantar flexion torque measured at −10° (r^2 = 0.84 and 0.75, respectively; $P<.01$), −20° (r^2 = 0.76 and 0.64, respectively; $P<.01$), and −30° (r^2 = 0.92 and 0.88, respectively; $P<.01$). MSIpas values were not significantly related with passive ankle plantar flexion torque values ($r^2 = 0.32$, $P>.05$) neither for −10°, −20°, nor −30°. MSIpas-neutral values were correlated with passive ankle plantar torque values measured at 0° ($r^2 = 0.47$, $P<.05$) (**FIGURE 5**) but not at −10°, −20°, and −30°. Finally, MSIpas values were not related with TSIpas, TSIpas-neutral, or TSIact values ($P>.05$), but MSIpas-neutral was correlated with TSIact ($r^2 = 0.45$, $P<.05$).

DISCUSSION

THE RESULTS OF THIS STUDY INDICATE that both the gastrocnemius muscle belly and Achilles tendon were elongated by passive stretching of the muscle-tendon unit performed with progressive ankle dorsiflexion. Lengthening of the Achilles tendon (approximately 3% of the initial length at 0°), although smaller in magnitude than that of the gastrocnemius muscle belly (approximately 10% of the initial length at 0°), was significant. Our findings of significant lengthening of the tendon under passive stretch are in agreement with the results of previously reported studies performed on animals[8,32,33] as well as humans.[10,23] The tendon lengthening determined in this study could have included tautening of the tendon slackened at the most plantar-flexed position, although the present tendon strains are much larger than the previously reported tendon slackness (negative strain).[23] The present muscle belly elongation might include stretches of fascicles and the proximal tendon and proximal and distal aponeuroses of the MG. However, because the proximal tendon is very short, it does not provide sizable length change.[21] In addition, we have

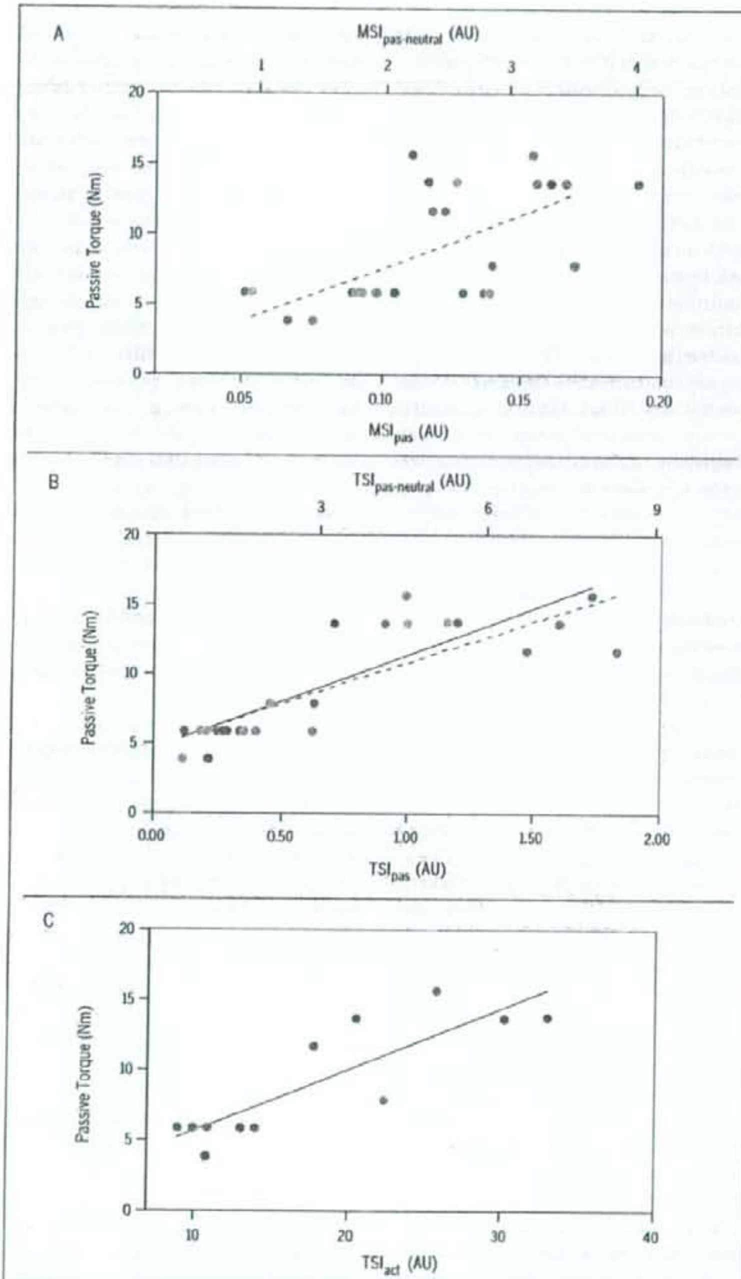

FIGURE 5. The relationship between passive torque and muscle belly stiffness index (MSIpas [bluez circles] and MSIpas-neutral [orange circles]) (top) and tendon stiffness index (TSIpas [blue circles] and TSIpas-neutral [orange circles], middle, and TSIact, bottom). In each panel, regression lines are also shown (top, MSIpas-neutral; middle, solid, TSIpas, dashed, TSIpas-neutral, bottom, TSIact).

previously established that there is no lengthening of the proximal aponeurosis of the MG by passive ankle dorsiflexion up to −30° of dorsiflexion.[21] However, the distal aponeurosis could possess compliance, especially in the passive condition,[23,38] and could have contributed to the muscle belly length change. Further research is needed to separate the MG muscle belly lengthening into those of fascicles and aponeurosis, but the very compliant fascicles in the passive condition should have significantly contributed to the muscle belly elongation in this study.

A novel finding of the study was the positive correlations between passive ankle plantar flexion torque and Achilles tendon passive and active stiffness indices. Highly significant correlations for the 2 indices of tendon stiffness strongly suggest that the extensibility of the Achilles tendon (distal to the MG muscle belly)[7] is one of the limiting factors to joint flexibility. For "tighter" subjects, the ankle position of 0° would be closer to the maximal range, as compared with "flexible" subjects. Thus it is possible that the TSIpas and TSIpas-neutral were affected by the nonlinear increase in torque as the ankle was dorsiflexed. However, because the MSIpas was not significantly correlated with the passive torque at 0°, −10°, −20°, and −30°, and because the TSIact separately measured at much higher force levels was related with the TSIpas and TSIpas-neutral and with the passive torque at 0°, −10°, −20°, and −30°, we consider our result to be not simply due to our calculation but to the association between tendon extensibility and passive torque. It has been suggested that joint flexibility is influenced by musculotendinous structures around the joint.[2,4,20,30,31,34] Gajdosik et al[4] concluded that the passive plantar flexion torque, which they defined as the flexibility of the ankle joint, is determined by the extensibility of musculotendinous structures—in particular, that of the gastrocnemius. Similarly, Riemann et al[26] suggested that the gastrocnemius muscle-tendon unit is a major contributor to passive joint stiffness. Both of them failed to separate muscle-tendon unit into respective components, and Gajdosik et al[8] speculated that the extensibility of muscle-tendon unit is predominantly determined by that of muscle fibers, and that tendinous tissues are too stiff to significantly contribute to lengthening. The present results do indicate that the Achilles tendon compliance does contribute significantly to passive ankle joint torque at 0°. The coefficient of determination (r^2) between TSIpas, TSIpas-neutral, and TSIact, and passive ankle joint torque at 0° ranged from 0.62 to 0.76 for the passive and active tendon stiffness indices, suggesting that about 70% of the variance of passive torque among subjects can be explained by the variance of tendon stiffness. Observation on rats has shown that the difference between compliant and stiff tendons affects the magnitude of elongation of a muscle-tendon unit at extreme joint positions.[9] In vivo studies on humans have shown substantial interindividual variability of tendon elasticity.[15] The present finding of the effect of tendon stiffness on the passive joint torque hints to the possibility that interindividual variability in joint mobility is regulated at least in part by the difference in tendon mechanical properties.

The disappearance of waviness in collagen fibers and their subsequent lengthening[24] is related with an exponential increase in passive torque by ankle dorsiflexion.[31] The tendon is rich in collagen fibers and, with its length sufficient for lengthening in the order of centimeters, could take up the elongation of muscle-tendon unit, reducing muscle fiber lengthening and passive torque. This would be the reason for the negative correlation between the tendon and muscle belly elongations in the passive condition. More compliant tendon might result in less elongation of the muscle belly for the same muscle-tendon unit length, resulting in a smaller passive torque.

The stiffness index of the gastrocnemius muscle belly at rest (MSIpas) was not correlated with the passive ankle plantar flexion torque. The MSIpas-neutral was related with the passive ankle plantar flexion torque at 0°, although the correlation was weaker compared with those for tendon stiffness indices. Significant correlation between one of the stiffness indices of the gastrocnemius muscle belly in a resting state and the passive ankle plantar flexion torque suggests the contribution of aponeuroses within the muscle belly to the resistance to stretch, as described above. This could explain the significant correlation between MSIpas-neutral and TSIact. Also, the connective tissues within the muscle fibers might contribute to the mechanical characteristics of a dissected muscle.[23] However, higher correlations for all of the stiffness indices of tendon with the passive ankle plantar flexion torque strongly suggest that interindividual variability of the mechanical characteristics of tendon is more closely related to joint flexibility. In fact, the coefficient of variation of tendon lengthening over subjects at 0° (44%) was much larger than that of the muscle belly (18%). Although the stiffness of muscle belly could have contributed to the passive ankle plantar flexion torque to some extent, the muscle belly might have been elongated just to a level primarily determined by the distal Achilles tendon elongation and the latter governed the passive torque at 0°. Stiffer tendon would lead to larger lengthening of muscle fibers for the same joint position, hence greater passive torque and more severe sensation of stretch and pain that limits further joint displacement. Stretch-induced activation of muscle fibers and accordingly an increase in muscle belly stiffness might be additional factors limiting muscle belly lengthening. This effect, if any, would have been minimal in this study, because care was taken not to provoke pain sensation from the subject in an attempt to avoid muscle activity. Therefore, the results suggest that the extensibility of the Achilles tendon is a major determinant of the joint passive torque of the ankle.

There are several imitations in the present study: (1) there was only a small number of heterogeneous subjects (females and males), some of whom did not complete all test angles, (2) the end range of joint motion was not investigated, and (3) only the gastrocnemius and Achilles tendon were studied. Regarding the first limitation, we pooled females and males in a single group. However, a significant linear relationship between tendon stiffness indices and passive torque strongly suggests that the association between the 2 parameters are similar for females and males. But there are known differences between sexes in joint range of motion that may have influenced the ability to establish the effect of muscle belly extensibility on passive joint torque. Future study on a large population of both sexes will reveal if this is the case. Regarding the second limitation, joint flexibility is commonly evaluated based on the maximal amount of range of motion, not by the passive torque at a fixed position within the range. But the 2 concepts are closely related,[30] and musculotendinous stiffness correlates negatively with the amount of joint motion.[35] Although further research is warranted, we consider that the end range of motion is related with passive torque at a position within the range, and that tendon stiffness similarly affects the range of motion. Regarding the third limitation, joint structures other than the muscle-tendon component (ie, ligaments and joint capsule) may have also contributed to the resistance to motion. We cannot rule out the possibility that the present tendon stiffness indices included resistance to motion provided by ligaments and joint capsule. But for the relatively low load at 0° (passive torque of 16 Nm in the tightest subject) and even for further dorsiflexion (-61 Nm), these structures would have contributed far less than the tendon. The TSIact could have included these structures, which might be the reason for the lower correlation between passive tendon stiffness indices (TSIpas and TSIpas-neutral) compared with that between TSIpas and TSIpas-neutral. It should also be noted that we assumed that the gastrocnemius muscle-tendon unit represented the behavior of musculotendinous structures during passive dorsiflexion. But other plantar flexor muscles, such as the soleus, would have contributed to the passive torque. However, the Achilles tendon is shared both by the gastrocnemii and soleus muscles. Therefore, we consider that the present tendon length change represented the triceps surae muscles. Also, there is a possibility of interindividual difference in the tolerance to uncomfortable stretch sensation that may have contributed to the joint mobility[3] or the inability of some individuals to complete all components of the assessment procedure. As described earlier, in the present study we discarded this possibility by minimizing muscular activity with the limited range of motion for assessment of passive torque and stiffness indices, avoiding pain sensation of the subjects.

The findings of the present study have some clinical implications. When clinicians evaluate ankle joint range of motion or passive torque within the range, they would be able to evaluate extensibility of the Achilles tendon. This opens the possibility of evaluation of the rehabilitation process after Achilles tendon injuries. Injuries frequently occur not at the extreme range of joint motion but in the middle of extremes,[35] which might be related in the case of the ankle with the interindividual differences in passive torque and Achilles tendon extensibility. It is possible that passive torque measurement could be used as a test for screening injury-prone populations. In addition, previous in vivo studies have shown that the tendon changes its viscoelastic properties by repeated contractions[15] and static stretching.[14] These changes in tendon material properties might be the basis for the preconditioning effect on the joint range of motion used by physical therapists, and studies are underway in our laboratory to clarify this mechanism.

CONCLUSION

THE PRESENT RESULTS SUGGEST THAT extensibility of the muscle-tendon unit—that of the Achilles tendon for the most part—is related with the passive ankle plantar flexion joint torque. This result has implications for ankle joint flexibility and could have clinical relevance in the field of musculotendinous injury prevention and rehabilitation.

KEY POINTS

FINDINGS: Passive resistance to ankle dorsiflexion is related to the stiffness of the Achilles tendon rather than the gastrocnemius muscle belly.

IMPLICATION: Stretching of calf muscles lengthens the Achilles tendon, the amount of which could be a factor of interindividual differences in ankle joint flexibility.

CAUTION: This study could have benefited from a larger number of subjects with diverse ankle joint range of motion.

ACKNOWLEDGEMENTS: *This work was partly supported by the Ministry of Education, Science, Sports and Culture, Grant-in-Aid for Scientific research 15500406, 2003-2005. The authors wish to thank to Toshiaki Oda, PhD (RIKEN [Japan]) and Kentaro Chino, PhD (Japan Institute of Sports Sciences) for their technical assistance.*

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Journal of Neuroscience Methods 173 (2008) 241–248

Contents lists available at ScienceDirect

Journal of Neuroscience Methods

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

Frequency features of mechanomyographic signals of human soleus muscle during quiet standing

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ARTICLE INFO

Article history:
Received 2 April 2008
Received in revised form 6 June 2008
Accepted 11 June 2008

Keywords:
Mechanomyogram
Postural sway
Recurring muscle activity
Cross-spectral analysis

ABSTRACT

The purpose of the present study was to determine whether the frequency features of the MMG signals during quiet standing reflect body sway as well as recurring muscle activity. Twenty healthy men maintained quiet standing in a barefoot position with their eyes open or closed. During quiet standing, MMG detected using uniaxial piezoresistive accelerometer and surface electromyogram (EMG) signals were recorded from the soleus (SOL) muscle, and the center of mass (CoM) displacement (CoMdis) in the anteroposterior direction was measured by a high-resolution laser displacement sensor. In addition, CoMdis was time-differentiated to yield CoM velocity (CoMvel). Cross-spectral analysis revealed that significant coherency spectra from MMG to CoMdis and from MMG to rectified EMG of SOL were observed below 2 Hz and 8–12 Hz frequency band, respectively. Furthermore, we revealed that the trajectories of MMG and the calculated dMMG/d*t* were significantly correlated to CoMdis and CoMvel, respectively. These results suggest that kinematic and physiological parameters of postural control during quiet standing can be quantified by frequency features of the MMG.

1. Introduction

Mechanomyogram (MMG) involves recording and quantifying the low-frequency oscillations produced by activated skeletal muscles (Barry, 1987; Frangioni et al., 1987). Studies using evoked contractions by electrical stimulation of isolated motor units in rat (Bichler and Celichowski, 2001a,b), cat (Orizio et al., 1999, 2000) and human (Yoshitake et al., 2002) gastrocnemius muscles provided evidence that MMG signals are dependent on the contractile properties of the activated motor units. In voluntary contractions in humans, many researchers have succeeded in investigating the amplitude and frequency responses of MMG signals during anisometric as well as isometric contractions under well-controlled conditions (reviewed by Beck et al. (2005) and Orizio (1993)). In addition, fatigue-related changes in the activation strategies and contractile properties of motor units have been assessed by means of MMG (reviewed by Shinohara and Søgaard (2006)). Although the fundamental findings of the MMG characteristics have been reported by many researchers, there are only a few research groups that have examined the functional performance of daily activities in relation to MMG; furthermore, the task investigated in these studies is limited to a bicycle exercise (Housh et al., 2000; Perry et al., 2001; Shinohara et al., 1997).

Upright standing is one of the most basic daily human activities, and postural stability during quiet standing deteriorates with age (Maki et al., 1990; Panzer et al., 1995) or inactivity (Kouzaki et al., 2007). The deterioration of equilibrium control is associated with an increased risk of falls in elderly persons (Gehelsen and Whaley, 1990), and, therefore, postural control during upright standing has strong functional significance in daily living. Based on the dynamics of the human quiet stance, it has been observed that the plantar flexor muscles play a significant role in stabilizing the body during quiet standing (Masani et al., 2003; Morasso and Schieppati, 1999). Additionally, the activities of the plantar flexors have been found to be coherent with both spontaneous body sway (Gatev et al., 1999; Masani et al., 2003) and mechanically induced body sway (Fitzpatrick et al., 1996). From these previous findings, therefore, in order to examine the human postural control mechanism revealed by MMG, a technique is needed that can detect and quantify not only the activities of the plantar flexor muscles but also the kinematic parameters of postural control.

The MMG signals include the displacement of moving parts of the body, not only the changes in the muscle itself (Watakabe et al., 2001). Up to now, the signals related to body or limb movements

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doi: [illegible]

have been regarded as artifacts, and therefore most studies have used a filter with a 5 Hz high pass cutoff frequency to attenuate movement artifacts in MMG signals (Beck et al., 2005). According to the model of a single joint inverted pendulum rotating around the ankle joint (Masani et al., 2003; Morasso and Schieppati, 1999), the low-frequency component of the MMG signal without a high pass cutoff filter is likely to represent the displacement of body sway during quiet standing.

In addition to the dynamics of the bipedal quiet stance as a postural control mechanism, an upright posture is partially stabilized by recurring muscle activity of the soleus (SOL) muscle. Mori (1973, 1975) reported that there is a motor unit synchronization recurring at around 10 Hz in the SOL during quiet standing in healthy subjects. Mochizuki et al. (2005) also observed evidence that different motor units were synchronized within the SOL during quiet standing. The recurring muscle activity around 10 Hz produced by motor unit synchronization appears as physical oscillations of the body surface on the activated muscle fibers (Lippold, 1970). These oscillations are referred to as physiological tremor (McAuley and Marsden, 2000). It has been reported that during sustained contraction of even low intensity, the involvement of the physiological tremor has a large influence on the MMG signals (Goldenberg et al., 1991; Orizio, 1993). Thus, the recurring muscle activity during quiet standing can be extracted by frequency-domain analysis of the MMG signals.

The literature indicating the frequency features of the MMG signals leads us to hypothesize that the MMG signals make it possible to evaluate kinematic and physiological parameters related to postural control mechanisms. To this end, in the present study, the frequency characteristics of the MMG were compared with body swaying and recurring muscle activity during quiet standing.

2. Methods

2.1. Subjects

Twenty young men (range: 23–35 years) volunteered for this experiment. They gave their written informed consent for the study after receiving a detailed explanation of the purposes, potential benefits, and risks associated with participation in the study. All subjects were healthy and had no history of any neurological disorders, and their vision was corrected to normal levels. All procedures used in this study were in accordance with the Declaration of Helsinki and were approved by the local ethical committee.

2.2. Experimental protocol and measurement

The basic procedure setup and measurement of postural sway during quiet standing has been described in our previous studies (Kouzaki et al., 2007; Masani et al., 2003, 2007). The subjects were required to maintain a quiet stance barefoot on a platform with their eyes open (EO) or closed (EC) and with a distance of 15 cm between their heels, for approximately 70 s. The subjects held their arms by their sides. Three trials were conducted for each eye condition, and sufficient resting time was allowed between trials. The order of the trials was pseudo-randomized.

A surface mechanomyogram (MMG) detected using an uniaxial piezoresistive accelerometer (ASV-2GA, Kyowa, Tokyo, Japan) was recorded from the muscle belly of the right soleus (SOL) muscle. We utilized the accelerometer and not a microphone to detect the body sway as well as muscle activity, because the accelerometer records the displacement of moving parts of the body better than the microphone (Watakabe et al., 2001). The accelerometer had a flat frequency response from DC to 150 Hz, and its physical dimensions were 22 mm times 22 mm for the base, 11 mm in height, and 13 g in mass. The MMG attached to the posterior part of the SOL so that it was as flat as possible within the SOL and so that the limb movement could be detected in the anteroposterior direction (Fig. 1). Furthermore, the accelerometer was secured over the muscle belly of the SOL with adhesive tape so that the polarization between positive and negative values represents the inward and outward directions from the surface, respectively. To minimize the influence of signals from the neighboring synergists (i.e., medial and lateral gastrocnemius) to SOL, the location of the accelerometer was determined using ultrasound B-mode images (SSD-900, Aloka, Tokyo, Japan), which can monitor visually the boundary between SOL and gastrocnemius muscles.

The surface electromyogram (EMG) of the right SOL located laterally near the MMG was recorded using bipolar Ag–AgCl electrodes with a diameter of 10 mm and an interelectrode distance of 20 mm. The electrodes were connected to a preamplifier and a differential amplifier (×1000) having a bandwidth of 20–500 Hz (SX203, Biometrics Ltd., Gwent, U.K.).

The reference electrode for the EMG was placed on the medial malleolus. To assess the trajectory of the center of mass (CoM) displacement (CoMdis), the horizontal position of a lumbar point at L3 was measured by a laser displacement sensor (1 μm resolution, LK-2500, Keyence, Osaka, Japan) (Masani et al., 2003) (Fig. 1). Laser displacement sensor makes it possible to detect the range of 30 to 40 cm from the source of luminescence. All electrical signals were stored with a sample frequency of 1 kHz by a 16-bit analog-to-digital converter (PowerLab/16SP, ADInstrument, Sydney, Australia) and stored on the hard disk of a personal computer for later analyses.

2.3. Data analysis

For all recorded signals, data for a 60 s period in the middle portion of the collected data (~70 s) were selected for analysis of individual trials.

The auto-spectral analysis was performed for the CoMdis, the full-wave rectified EMG of the SOL, and the MMG of the SOL. For frequency characteristics of surface EMG, we adopted the rectified EMG because rectification of the surface EMG signal and its power spectrum is a strategy that has been used to reveal the temporal pattern of grouped motor unit discharges (Halliday et al., 1995; Myers et al., 2003), and the activation strategy of the muscles (Yoshitake et al., 2007). The cross-spectral analysis from MMG to EMG and CoM was performed to investigate the coherency and time shift between MMG signals and the muscle activity and body sway. These frequency domain analyses were executed according to Bloomfield (2000). The data for 60 s were divided into 13 subsets with a length of 8192 data points (8.192 s). Almost half of the segment overlapped with the adjacent segments. A 13-bit fast-Fourier transform algorithm was then applied to generate a periodogram for each subset. Consequently, the frequency resolution was 0.122 Hz. An ensemble-average auto-power spectral density was calculated across these segments.

The coherency spectrum $[\mathrm{Coh}^2(f)]$ for the two time series (x and y) was given as follows:

$$\mathrm{Coh}^2(f) = \frac{|S_{xy}(f)|^2}{S_x(f)S_y(f)}$$

where f denotes frequency. $S_{xy}(f)$ is the cross-power spectrum function of x and y, $S_x(f)$ and $S_y(f)$ are the auto-power spectrum functions of x and y, respectively. The phase spectrum $[\theta_{xy}(f)]$ was defined as:

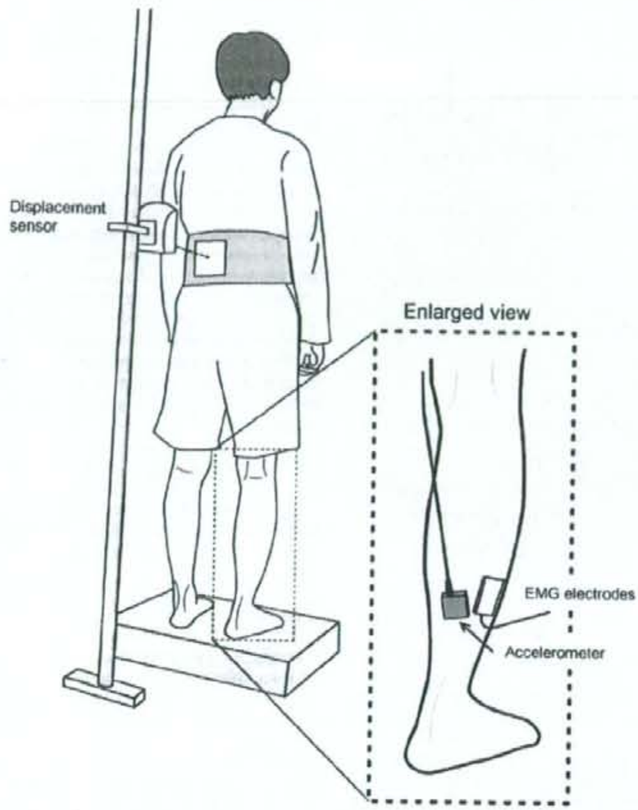

Fig. 1. Diagram of the experimental setup for measurement of the displacement of body sway, the mechanomyogram (MMG), and the surface electromyogram (EMG).

$$\theta_{xy}(f) = \tan^{-1}\frac{\cos S_{xy}(f)}{\sin S_{xy}(f)}$$

The statistically significant level at 1% for the coherency corresponded to 0.165 in this procedure. When the phase plotted against frequency at the coherent frequencies, the linear slope represents a constant time lag between the two signals (McAuley et al., 1997). In this case, the phase shift and the frequency have the following relationship:

$$\theta_{xy}(f) = 2\pi f\tau$$

Thus, we calculated the time shift as the value of the linear regression slope of the phase plot ($2\pi\tau$) divided by 2π.

To examine the relationships between postural sway and MMG signals, the body sway component was extracted using the MMG signal. First, the time series for the CoMdis was low-pass filtered with a cutoff frequency of 4 Hz (Masani et al., 2003) using a 4th ordered Butterworth filter employing a zero-phase lag. The filtered CoMdis was time-differentiated to yield CoM velocity (CoMvel). In addition, MMG signals were also low-pass filtered at a cut-off frequency of 1 Hz, because highly significant coherency from MMG and CoMdis was obtained below 1 Hz (see section 3) and a large fraction of the signals concerning body movements has been reported in the frequency range below 1 Hz (Fitzpatrick et al., 1992; Kouzaki et al., 2007; Masani et al., 2003). Then filtered MMG was time-differentiated to calculate dMMG/dt.

2.4. Test for very low-frequency characteristics of MMG during movement artifact

Previous study by Watakabe et al. (2001) has demonstrated that the MMG signals by accelerometer include the displacement of moving parts of the body, not only the changes in the muscle itself. Therefore, it is natural that low-pass filtered MMG represents displacement of the limb. However, there was no experimental report about it. Therefore, a supplemental test was conducted to examine the very low-frequency component of the MMG signal by accelerometer because the task employed in the present study includes body or limb movements as well as muscle activity. We measured the MMG signals by accelerometer during sinusoidal sway of accelerometer at a constant displacement to examine whether signals obtained from accelerometer are influence by slow cyclic change in displacement of the accelerometer. Diagram of experimental setup for mechanical sinusoidal vibration test is illustrated in Fig. 2a. The mechanical vibrator (DPS-285, Dia Medical System, Tokyo, Japan) oscillated the wooden bar attaching the accelerometer at constant frequency of 1 Hz. The displacement

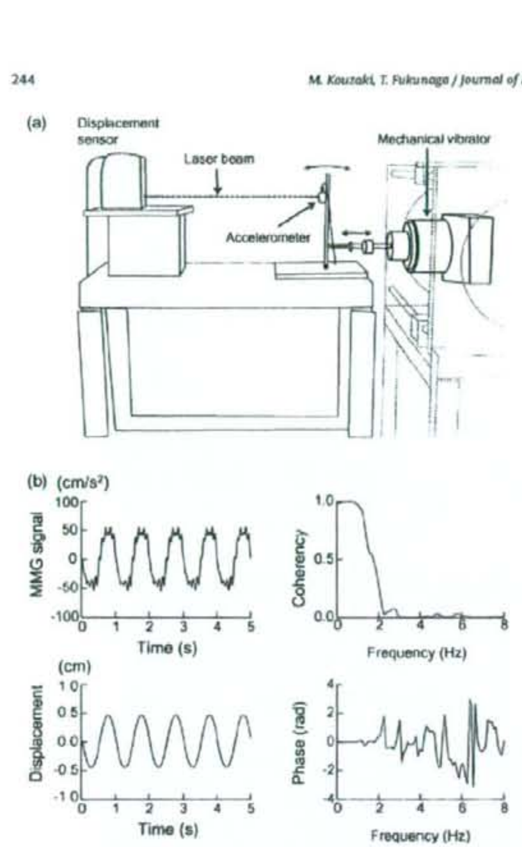

Fig. 2. Schematic drawing of the test for low-frequency characteristics of MMG signal during mechanical sinusoidal vibration (a). Wooden bar attaching the accelerometer sways in inverted pendulum manner. The wooden bar is oscillated by mechanical vibrator (DPS-285, Dia Medical System, Tokyo, Japan), which fixes to the metal frame of a metal stand supported by the ground. The mechanical vibrator consists of a direct-current motor with the shaft embedded in spring to remove the transmission of mechanical noise to wooden bar. Results of mechanical sinusoidal vibration test (b). Time series of MMG signal from an accelerometer and displacement of the accelerometer during sinusoidal sway of 1 Hz (*left column*). Coherency and phase spectra from the displacement of the accelerometer to the MMG signal (*right column*).

of accelerometer was measured by the high-resolution displacement sensor (LK-2500, Keyence, Osaka, Japan). Vibration frequency, pattern, and amplitude are controlled by the function generator (AFG-3022, Tektronix, OR, U.S.A.). The amplitude of vibrator was fine regulated so that the sway amplitude of the accelerometer in a horizontal direction is ± 0.5 cm. The mechanical sinusoidal vibration was performed for approximately 30 s. All electrical signals were stored with a sample frequency of 100 Hz, and data for a 20.48 s (2048 data points) period were selected for cross-spectral analysis. An 11-bit fast-Fourier transform algorithm was applied to generate a periodogram, and then coherency and phase spectra from the displacement of the accelerometer to MMG signals were calculated in similar procedure.

MMG signals by accelerometer and displacement of accelerometer during low-frequency sinusoidal sway are demonstrated in left panels of Fig. 2b. It seems likely that MMG signals have a great content of displacement. Cross-spectral analysis between MMG signals and displacement revealed that high coherency was observed up to around 1 Hz without phase difference. This finding experimentally revealed that the low-frequency component of MMG signals represents the displacement. In the present study, the low-pass filtered MMG was expressed as arbitrary unit (a.u.) because low-pass filtered MMG during motion artifact does not absolutely express the displacement.

On the basis of mathematical theory, displacement should be calculated by the double integral of MMG signal as acceleration. However, the displacement obtained from double integral of acceleration includes greatly integral error. During human behavior without well-controlled conditions, signals obtained from accelerometer contain various noises relating body or limb movements. Thus, the acceleration signals at low-frequency component are usually removed by high-pass filter procedure (Orizio, 1993) when the displacement is calculated by double integral of acceleration signals. An experimental design in the present study focuses on body or limb movement as well as muscle activities. Therefore, present study did not adopt the displacement calculated by the double integral of MMG as acceleration.

2.5. *Statistical analyses*

Linear regression analysis was conducted between CoMdis and MMG and between CoMvel and dMMG/d*t*. Data are given as means $\pm$ S.D. in the text and table, and as means $\pm$ S.E. in the figures.

3. Results

Firstly, in order to examine frequency-domain association between different signals, auto-spectral and cross-spectral analyses were performed for CoMdis, rectified EMG and MMG signals without filtering procedure.

A typical time series of the CoMdis, the rectified EMG of SOL and the MMG of SOL during quiet standing in one subject is shown on the graphs on the left in Fig. 3. It can be seen that the slow component of the MMG of SOL is similar to the fluctuations in CoMdis while the EMG of SOL exhibits tonic activity in a global view. The graphs on the right in Fig. 3 show the auto-power spectral density of the corresponding CoMdis, the rectified EMG of SOL and the MMG of SOL measurements. The power components of CoMdis and MMG of SOL monotonically decreased with increasing frequency, and a major part of the power was in the frequency range below 1 Hz. For the rectified EMG of SOL, a prominent component was observed at a frequency around 10 Hz.

3.1. *Cross-spectral analysis from MMG to postural sway and muscle activity*

An example of the cross-spectral analysis from the MMG of SOL to CoMdis for one subject is shown on the graphs on the left in Fig. 4. There was a significant coherency up to around 2 Hz with high coherency within 1 Hz. The phase spectrum from MMG to CoMdis was plotted only for the frequency at which the significant coherency was observed. The phase spectrum for the significant frequency range was distributed about zero, indicating no time delay for this pair. The right graphs of Fig. 4 show the cross-spectral analysis between the EMG and MMG of SOL for one subject. A significant coherency was observed around 10 Hz. The phase spectrum continued with a slope around 10 Hz.

The group averaged values of the results obtained from cross-spectral analysis are indicated in Fig. 5. There were significant coherency spectra from the MMG of SOL to CoMdis up to 2 Hz (EO:

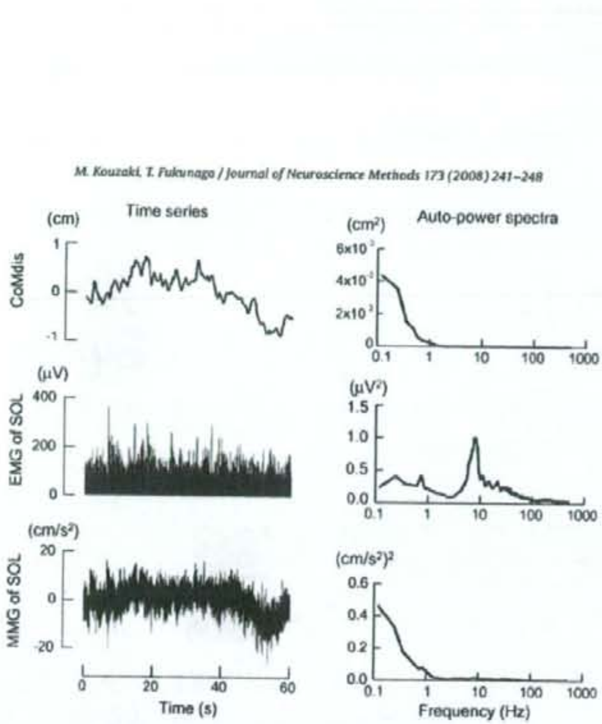

Fig. 3. Representative example of time series of center of mass (CoM) displacement (CoMdis), rectified electromyogram (EMG) of the soleus (SOL), and mechanomyogram (MMG) of the SOL (*left column*) without filtering procedure for a single trial during quiet standing, and the corresponding auto-power spectra (*right column*). From the top, the time series and auto-power spectra for CoMdis, the EMG of SOL, and the MMG of SOL. Note that the horizontal axes of the graphs on the right are presented in logarithmic scale to show the entire spectra.

1.9 ± 0.1 Hz, EC: 2.1 ± 0.1 Hz) during quiet standing. The time shift calculated from the slope of the phase spectra was considerably close to zero for both eye conditions. A significant coherency from the EMG to the MMG of SOL was observed in the frequency ranging from 8 to 12 Hz (EO: 8.7–11.8 Hz, EC: 7.8–11.9 Hz). The time shift corresponded to approximately 0.020 ± 0.011 s and 0.022 ± 0.010 s under the EO and EC conditions, respectively.

3.2. Association between MMG and postural sway

To examine the association between MMG signals and postural sway, MMG signals were low-pass filtered at a cutoff frequency of 1 Hz, because highly significant coherency from MMG and CoMdis was obtained below 1 Hz (Figs. 4 and 5). Typical time series of the CoMdis, CoMvel, filtered MMGs (cutoff frequency = 1 Hz), and

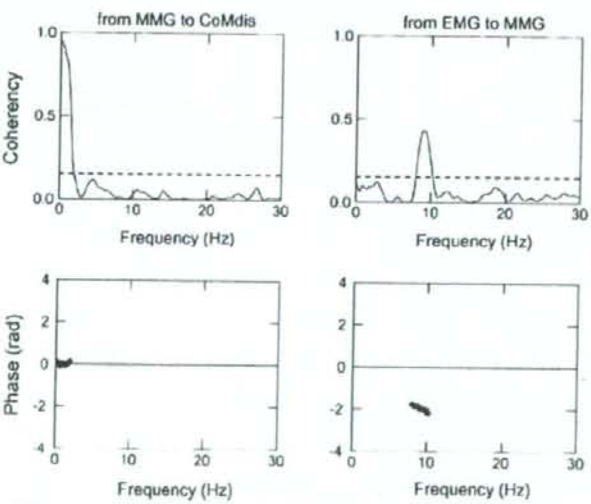

Fig. 4. Results of the cross-spectral analysis from the MMG of SOL to CoMdis (*left column*), and from the EMG of SOL to the MMG of SOL (*right column*) for one subject. The upper and lower panels indicate coherency and phase spectrum, respectively. The broken line represents the 1% significance level. The phase spectrum was plotted only for the frequencies that have significant coherency.

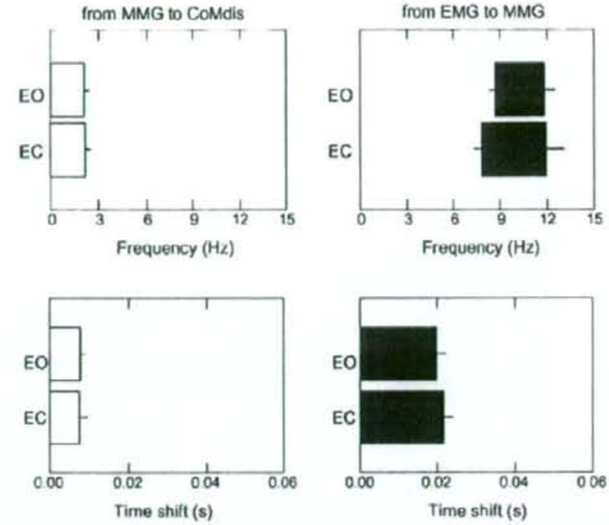

Fig. 5. Group-averaged frequency region (*upper panels*) and time shift (*lower panels*) where the significant coherency was observed from the MMG of SOL to CoMdis (*left column*) and from the EMG of SOL to the MMG of SOL (*right column*). EO and EC indicate the eyes-open and eyes-closed conditions, respectively.

dMMG/*dt* for a single subject during quiet standing are illustrated in Fig. 6a. It seems that the trajectories of CoMdis are similar to those of filtered MMG, which employ the cutoff frequency of 1 Hz. Likewise, it appeared that the CoMvel coincided with dMMG/dt. Fig. 6b shows the relationships between the body-sway parameters and MMG signals during quiet standing from the time series in Fig. 6a. The trajectories reflecting the relationship between CoMdis and filtered MMG were positively correlated. Similar to this relation, positive relationships were evident from the trajectories of the CoMvel and dMMG/dt. Table 1 summarizes the results for the correlation coefficient of those linear relationships as group averaged values. Highly positive values for the correlation coefficient were observed in the relations between CoMdis and filtered MMG, and between CoMvel and dMMG/*dt* in both eye conditions.

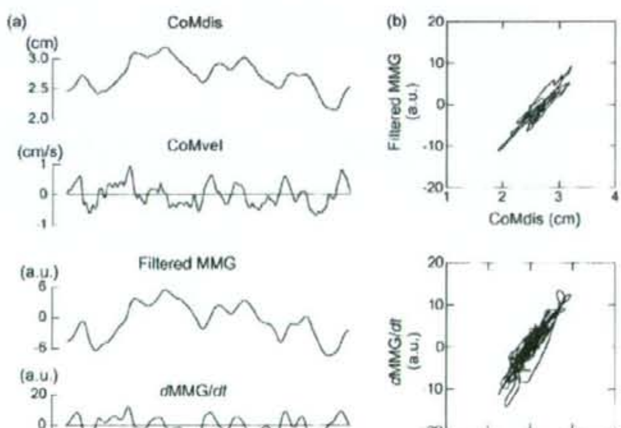

Fig. 6. Typical example of CoMdis, CoM velocity (CoMvel), filtered MMG of the SOL, and dMMG/*dt* (a), and the relationships between CoMdis and MMG, and between CoMvel and dMMG/*dt* (b). The time series of CoMdis and MMG were passed through a low-pass filter of 4 Hz and 1 Hz using a 4th-order Butterworth filter, respectively. See text for details.

Table 1
Summary of the results of the correlation coefficient between center of mass (CoM) displacement (CoMdis) and filtered mechanomyogram (MMG), and between CoM velocity (CoMvel) and dMMG/dt

	EO	EC
CoMdis vs. filtered MMG	0.880 ± 0.080	0.908 ± 0.073
CoMvel vs. dMMG/dt	0.808 ± 0.085	0.866 ± 0.066

EO and EC indicate eyes open and closed conditions, respectively.
Mean ± S.D.

4. Discussion

The main findings of this study were that: (1) a significant coherency from MMG to CoMdis and from MMG to the muscle activity of the SOL was observed in bands below 2 Hz and from 8–12 Hz during quiet standing, respectively; and (2) the trajectories of MMG and calculated dMMG/*dt* were significantly correlated to CoMdis and CoMvel, respectively. To our knowledge, this is the first paper examining the MMG during upright standing in humans. Our novel finding was that kinematic and physiological factors as related to the postural control mechanism were revealed by frequency features of MMG.

4.1. Association between MMG and body sway

The bipedal upright stance is inherently unstable because in this stance a large body mass with a high elevation center is kept in an erect posture over a relatively small base of support. According to a single inverted pendulum model, the ankle joint has the primary role for maintaining CoM equilibrium with respect to the supporting surface (Fitzpatrick et al., 1992, 1996; Gatev et al., 1999; Masani et al., 2003). Plantar flexor activity during quiet stance is a major determinant of ankle joint torque because the CoM is in front of the ankle joint (Smith, 1957), and the tibialis anterior muscle as an antagonist maintains a strict silence (Gatev et al., 1999; Kouzaki et al., 2007; Masani et al., 2003). Additionally, the activities of the plantar flexors have been found to be proportional to CoM trajectory (Gatev et al., 1999; Masani et al., 2003), and the muscle activities of plantar flexors preceded CoM fluctuations (Masani et al., 2003). These previous findings suggest that the plantar flexors control CoMdis via neural regulation to maintain an erect posture. Therefore, CoMdis is a significant parameter for examining the postural control mechanism.

We tried to assess the CoMdis by means of MMG because the MMG signals include the displacement of moving parts of the body, not only the changes in the muscle itself (Watakabe et al., 2001). So far, the signals relating to body movements have been regarded as an artifact, and therefore, most studies have used a filter with an appropriate high pass cutoff frequency to remove the movement artifact (Beck et al., 2005). We focused on the low-frequency component of the MMG signals for detecting CoM fluctuations, and compared the MMG with CoMdis by cross-spectral analysis. According to the qualitative observation shown in Fig. 1, the slow component of the MMG trajectories is similar to the CoMdis fluctuations. The cross-spectral analysis demonstrated the coherency from the MMG of the SOL to CoMdis below approximately 2 Hz without a time lag, suggesting that the very low frequency component of the MMG signals includes the changes in CoMdis during quiet standing. During quiet standing, the sway angle is less than 1 deg (Winter et al., 1998), and, therefore, the sway range in the anteroposterior direction of the MMG sensor corresponds to below 2 mm if the sensor is located approximately 10 cm above the ankle joint. Despite this imperceptible change, the MMG was found to be coherent with the CoMdis. Furthermore, there was a highly positive correlation between low pass-filtered MMG and CoMdis. These results, therefore, demonstrate that the MMG using an accelerometer employed in the present study makes it possible to evaluate the small and complex variability of CoMdis during quiet standing.

Morasso and Schieppati (1999) suggested that the process in the central nervous system that integrates multisensory information to obtain position and velocity information of the CoM is needed to stabilize the erect posture in computational simulation. In a study similar to the theoretical study, Masani et al. (2003) experimentally provided evidence that the velocity information of the CoM makes a crucial contribution to the control of the quiet stance. We therefore compared the calculated dMMG/*dt* with CoMvel to examine whether the velocity information during quiet standing is extracted from the MMG signals. As a result, the trajectories of dMMG/*dt* were linearly correlated with those of the CoMvel during quiet standing. Thus, integrating the present results concerning the association between the MMG signals and body sway suggests that the very low component of the MMG signal and its differentiation (dMMG/*dt*) reflect CoMdis and CoMvel as kinematic parameters of postural sway during quiet standing, respectively.

4.2. MMG and recurring muscle activity

It has been reported that the recurring muscle activity of the SOL during quiet standing resulted from the existence of the synchronization of motor units at around 10 Hz (Mochizuki et al., 2005; Mori, 1973, 1975). The recurring muscle activity around 10 Hz originating from motor unit synchronization appears as the physical oscillation of the body surface on the activated muscle fibers, which has been widely termed physiological tremor (McAuley and Marsden, 2000). Many researchers have suggested that physiological tremor is partly due to rhythmic modulation of the activity of several motor units caused by a servo-loop oscillation in the stretch reflex arc as a result of muscle spindle activity (Freund, 1983; Kouzaki et al., 2004; Lippold, 1970). It has been suggested that the continuously recurring activity of the SOL stabilizes the upright stance (Fitzpatrick et al., 1992). When muscle fiber does not stretch during quiet standing, Ia afferents are active as a result of α-γ linkage (Kouzaki et al., 2000). Furthermore, it has been pointed out that enhanced ankle stiffness produced by tonic muscle contraction of the plantar flexors provides partial compensation for the effect of gravity during quiet standing (Loram and Lakie, 2002; Morasso and Sanguineti, 2002; Morasso and Schieppati, 1999). Therefore, the recurring muscle activity of the SOL observed during quiet standing is a significant parameter for contributing erect posture.

In MMG studies in the past, the tasks have been designed to investigate the contractile properties of the muscles during fatigue for the sake of avoiding tremors (Barry et al., 1985; Kouzaki et al., 1999), since the involvement of the fatigue-related physiological tremor has a large influence on the MMG signals (Goldenberg et al., 1991; Orizio, 1993). We tried to examine whether the MMG can extract the recurring muscle activity even during quiet standing without fatigue. Cross-spectral analysis revealed that a significant coherency from rectified EMG to MMG is observed in the frequency range of 8–12 Hz, suggesting that the MMG signal makes it possible to detect the recurring muscle activity of the SOL even during static standing. In other words, the high frequency component of the MMG signal predominantly represents the recurring muscle activity of the muscle in the case of bipedal quiet standing. The time shift calculated by the phase spectrum for the frequencies that have significant coherency corresponded to approximately 0.02 s under both the EO and EC conditions. This indicates that changes in EMG precede those in MMG at around 10 Hz by a time of approximately 0.02 s. It is natural that a time delay was obtained between the EMG and MMG, because the EMG and the MMG reflect the electrical and mechanical activities of the muscle fiber, respectively (Orizio, 1993). In the present study, the time delay over which the MMG changes after the responses of the EMG would correspond to the physiological range for the force generation process during voluntary contraction of the lower limb (van Ingen Schenau et al., 1995). Taking this into account, the high frequency component of the MMG signals at around 10 Hz represents the recurring muscle activity, which is one of the significant factors in the postural control mechanism.

In conclusion, the present findings provide the first evidence of the MMG signals during quiet standing as related to human daily activities. The results obtained regarding the association between the MMG and the kinematic and physiological parameters of postural control suggest that the frequency features of MMG allow it to represent CoMdis, CoMvel, and the recurring muscle activity of SOL during quiet standing.

Acknowledgments

This work was supported, in part, by a grant from the Descente and Ishimoto memorial foundation for the promotion of sports science (to M. K.).

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