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RESEARCH ARTICLES

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The role of plantar cutaneous sensation in unperturbed stance

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Abstract Considerable evidence shows that sensation from the feet and ankles is important for standing balance control. It remains unclear, however, to what extent specific foot and ankle sensory systems are involved. This study focused on the role of plantar cutaneous sensation in quasi-static balance control. Iontophoretic delivery of anesthesia was used to reduce the sensitivity of the forefoot soles. In a follow-up experiment, subjects received intradermal injections of local anesthetic into the entire weight-bearing surface of the foot soles. Properties of the center-of-foot-pressure (COP) trajectories and ground reaction shear forces were analyzed using stabilogram–diffusion analysis and summary statistics. Effects of foot-sole anesthesia were generally small and mostly manifested as increases in COP velocity. Magnitude of COP displacement was unaffected by foot-sole anesthesia. Forefoot anesthesia mainly influenced mediolateral posture control, whereas complete foot-sole anesthesia had an impact on anteroposterior control. During bipedal stance, statistically significant effects of foot-sole anesthesia on COP were present only under eyes-closed conditions and included increases in COP velocity (11–12%) and shear force root-mean-square (13%), the latter indicating increases in body center-of-mass accelerations due to the foot-sole anesthesia. Similar effects were seen for unipedal stance in addition to an increase in anteroposterior COP median frequency (36%). Changes in stabilogram–diffusion parameters were confined to the short-term region suggesting that sensory information from the foot soles is mainly used to set a relevant background muscle activity for a given posture and support surface characteristic, and consequently is of little importance for feedback control during unperturbed stance. In general, this study demonstrates that plantar sensation is of moderate importance for the maintenance of normal standing balance when the postural control system is challenged by unipedal stance or by closing of the eyes. The impact of reduced plantar sensitivity on postural control is expected to increase with the loss of additional sensory modalities such as the concomitant proprioceptive deficits commonly associated with peripheral neuropathies.

Keywords Posture · Balance · Fractal Brownian motion · Somatosensory · Foot

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Introduction

It has been estimated that 4.5 million Americans (including 20% of the elderly population) may be suffering from peripheral neuropathies, largely as a consequence of diabetes (Richardson and Ashton-Miller 1996; Apfel 1999). Peripheral neuropathy patients exhibit decreased stability while standing (Geurts et al. 1992; Simoneau et al. 1994; Boucher et al. 1995) as well as when subjected to posterior translations of the support surface (Inglis et al. 1994). Epidemiological evidence links peripheral neuropathies to an increased risk of falling (Richardson et al. 1992; Richardson and Hurvitz 1995). Since the patients in these studies were restricted to those without measurable motor deficits, it is apparent that sensory information from the periphery (i.e., shanks, ankles, and feet) is important for the maintenance of upright balance. It remains unclear, however, to what extent specific foot and ankle sensory systems are involved in balance control.

The balance deficits associated with peripheral neuropathy may be related to the reduced ankle position sensation

(Van den Bosch et al. 1995; Simoneau et al. 1996; van Deursen et al. 1998) and/or to reduced rate of ankle torque production (Gutierrez et al. 2001) seen in these patients. However, reduced plantar sensitivity is typically an earlier and more obvious sign of peripheral sensory neuropathy. Plantar cutaneous afferents could potentially provide valuable feedback to the balance control system regarding the production of ankle torque, weight transfer between the legs, the rate of limb loading, and/or the nature of the support surface. Perception of forces under the feet, tangential (shear direction) and perpendicular (normal direction) to the skin during stance, might also be used to generate an internal estimate of the body center-of-mass (COM) location (Morasso and Schieppati 1999). It is therefore reasonable to hypothesize that reduced plantar pressure sensitivity may be an important contributor to the balance deficits seen in peripheral neuropathy patients.

Altered plantar pressure sensation may contribute to the balance deficits seen in other populations as well. When compared with younger (50–60 years old) controls, elderly subjects (>85 years) exhibit greater sway velocities that may be related in part to somatosensory deficits in the lower legs and feet (Pyykko et al. 1990). Astronauts returning from prolonged exposure to microgravity exhibit increases in 'tremor-type' postural sway frequencies (Cherepakhin et al. 1973; Kozlovskaya et al. 1981) that might be due in part to hypersensitivity of the cutaneous foot soles (Cherepakhin et al. 1973; Kozlovskaya et al. 1981; Kozlovskaya et al. 1988). This notion is supported by the observation that periodic application of negative pressures to the lower limbs can mitigate the balance deficits associated with prolonged bed rest (Dupui et al. 1992).

Despite the association between reduced plantar sensitivity and balance difficulties, previous research has largely failed to determine the role of foot-sole sensation in standing balance. For most previously employed techniques, plantar cutaneous sensation remained unaltered or was not isolated from foot and ankle proprioception. Rotation of the support surface concurrent with body orientation ('sway referencing'; Nashner et al. 1982) or support surface translation ('nulled ankle input'; Bloem et al. 2000) has been used to reduce stimulation of the ankle joint proprioceptors. Under these circumstances, however, foot-sole sensitivity remains intact and may still provide information regarding the orientation of the support surface with respect to the gravity vector. Previous studies employing hypothermic (Magnusson et al. 1990a, 1990b), ischemic, (Mauritz and Dietz 1980; Hayashi et al. 1988; Horak et al. 1990) or pharmaceutical (Konradsen et al. 1993) anesthesia of the feet and ankles have largely failed to isolate the cutaneous soles. Notable exceptions to this general rule include the study of compensatory stepping responses by Perry et al. (2000) and brief tests of unperturbed stance by Asai et al. (1990), each involving hypothermia of the soles. Alternatively, high-frequency (Kavounoudias et al. 1999, 2001) or low-frequency (Maurer et al. 2001) vibrations of the foot soles have been used to manipulate plantar sensation. With each of these techniques, however, the resulting effects on standing balance could still be related to activation of intrinsic foot proprioceptors or protective reflexes as well as balance-related feedback from plantar pressure receptors. The present study is an attempt to isolate the role played by plantar cutaneous mechanoreceptors in the maintenance of unperturbed stance. Subjects underwent specific anesthesia procedures that targeted the plantar skin, eliminating confounding effects on other afferent and efferent systems within the feet. The first balance experiment investigated the effects of cutaneous anesthesia of the forefoot soles in healthy subjects (procedure 1). A follow-up experiment investigated the effect of whole-sole anesthesia on unperturbed stance (procedure 2).

Methods

The Boston University Charles River Campus Institutional Review Board approved the procedures in this study and all subjects provided informed consent. Participants reported no history of neurological disorder, cardiac arrhythmia, or sensitivity to lidocaine. As an inclusion criterion, all subjects were able to stand with eyes closed and with bare feet placed heel-to-toe for 30 s. Ten healthy subjects (five males and five females) aged 21 to 46 years (mean ±SD, 28.6±8.5) participated in the first experiment. Six healthy male subjects aged 19 to 46 (26±10) years participated in the second experiment. One subject participated in both experiments.

Procedure 1: reduction of plantar forefoot sensitivity

Anesthesia of the skin overlying the metatarsal heads of both feet was produced using alternating-pulse iontophoretic delivery of an anesthetic solution (3.4% lidocaine and 1:50,000 epinephrine; Fig. 1A). This method targets the end-organs of the cutaneous receptors while leaving muscle and joint afferents intact. Details of the iontophoretic anesthesia procedure have been reported elsewhere (Meyer 2003; Meyer and Oddsson 2003). Cutaneous pressure sensory thresholds were determined before and after anesthesia, and again after the completion of balance testing, using Semmes-Weinstein nylon monofilaments (Stoeling Co., Wood Dale, IL, USA). These filaments are calibrated such that a specific longitudinal force makes them buckle. The 5.07 a.u. (≈115 mN)[1] sensory threshold level is normally indicative of peripheral sensory neuropathy (Holewski et al. 1988; Sosenko et al. 1990; Vinik et al. 1995). To assess whether the anesthesia procedure had an effect on muscle proprioceptors, a test of toe position perception was performed.

Balance tests under control and anesthetized conditions were separated by up to 7 days, with half of the subjects performing the control tests on the first day. Under both foot-sole sensory conditions, subjects completed twelve 35-s trials while performing each of three tasks: bipedal stance with eyes closed (bipedal-EC), bipedal stance with eyes open (bipedal-EO), and unipedal stance with eyes open (unipedal-EO). Tasks were alternated to distribute any effects of fatigue equally. Subjects remained standing but were free to move for 45 s between trials and sat down for 5 min after every block of ten trials.

Ground reaction forces during stance trials were measured using a Kistler 9284 multi-component force platform, sampled at 100 Hz; data were stored on computer disk for later processing. The time-dependent point of application of the resultant vertical ground

[1] Pressure sensory threshold levels have historically been expressed in arbitrary units (a.u.) that are considered a linear scale of perceived force and correspond approximately to a logarithmic scale of filament buckling force.

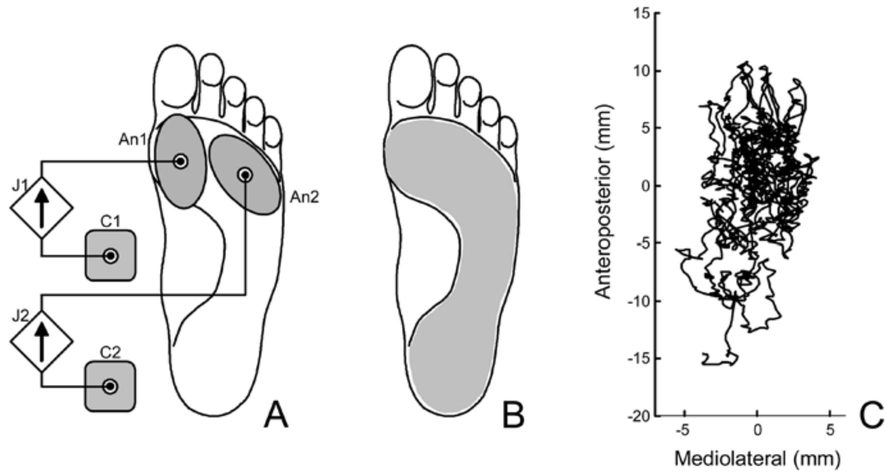

Fig. 1A–C Bilateral manipulations of plantar cutaneous sensation (left foot shown). **A** Placement of anodes *An1*, *An2* for iontophoretic delivery of a local anesthetic (procedure 1). Anti-phase currents are supplied by stimulus isolators *J1* and *J2*. Cathodes *C1*, *C2* are placed on the ipsilateral leg. **B** Anesthetized area of foot sole after intradermal injection of local anesthesia (procedure 2). **C** Example of 30-s center-of-pressure stabilogram (normal plantar sensitivity)

reaction force, or center-of-pressure (COP), was calculated from the ground reaction forces (see Fig. 1C for sample data). Subjects were instructed to stand upright on the force platform and to remain as still as possible without becoming stiff. They received verbal encouragement in this regard after each trial and viewed the COP time series from the previous trial on a computer monitor. During bipedal trials, subjects assumed a standardized stance (feet abducted 10° and heels separated mediolaterally by 6 cm (Collins and De Luca 1995). During unipedal stance trials, the dorsal surface of the unweighted, non-dominant foot was held in contact with the calf of the stance leg. If the subject's unweighted foot touched the floor the trial was repeated after a short rest period. For trials with eyes open, subjects gazed on a black 'X' placed at eye level 2 m in front of them. All other central and peripheral visual targets were removed. Templates were used prior to the start of each trial to standardize placement of the feet.

Post-processing of data was carried out using customized software (Matlab R.11; The Mathworks, Natick, MA, USA). COP data were filtered using a fourth-order zero-phase low-pass Butterworth digital filter with a cutoff frequency of 15 Hz. The initial and final 2.5 s of data were discarded to eliminate transient effects of the filter and 'settling time' in the subject's behavior. Postural activity was analyzed using the stabilogram–diffusion method (Roy et al. 1985; Collins and De Luca 1993), which involves modeling of COP trajectories as a series of correlated quasi-random walks. The analysis reveals two types of behavior of the COP where short time intervals ($\Delta t <\sim 1$ s) are dominated by positively correlated COP displacements (persistence), while longer time intervals (~ 1 s $<\Delta t <\sim 10$ s) are dominated by negatively correlated displacements (antipersistence). The intersection of the short- and long-term behavioral regions within the stabilogram–diffusion function (transition point, coordinates [Tt, Td]) was selected as the first point at which the slope of the function fell below unity (cf. Fig. 2). Effective diffusion coefficients (D, quantifying the degree of stochastic activity) and scaling exponents (H, describing the degree of correlation between successive COP displacements) were estimated for the short- and long-term regions along the anteroposterior (AP), mediolateral (ML), and planar (P) dimensions (vector sum of AP and ML dimensions). Since the transition from the long-term behavioral region to the saturation region (Chow and Collins 1995) occurs at shorter time intervals in unipedal stance than in bipedal, the long-term region regressions were performed over the

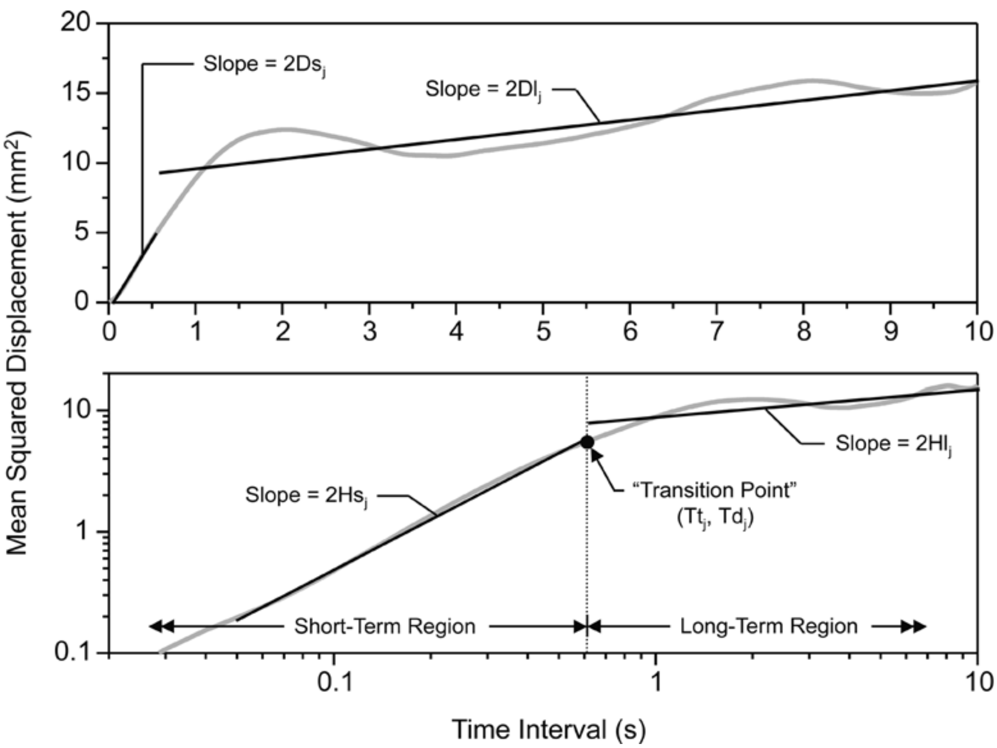

Fig. 2 Linear and log–log representations of the stabilogram–diffusion function $[\langle \Delta j^2 \rangle = f(\Delta t)]$ calculated from real center-of-pressure (COP) data are shown in *gray*. The straight lines in *black*, fitted to the short- and long-term regions of the function, are used to extract the different COP stabilogram–diffusion parameters. The *Transition Point* (Tt_j, Td_j) is the point where the stabilogram-diffusion function becomes antipersistent (scaling exponent H<0.5). Effective diffusion coefficients Ds_j and Dl_j are determined from linear regressions of the linear stabilogram–diffusion function over the ranges [0.05 s Tt_j] and [Tt_j 10 s], respectively. Likewise, scaling exponents Hs_j and Hl_j are determined from linear regressions of the log–log stabilogram–diffusion function over the ranges [0.05 s Tt_j] and [Tt_j 10 s], respectively

time interval range [Tt, 5 s] rather than [Tt, 10 s] that was used for bipedal stance (for details on the stabilogram–diffusion analysis procedure see Meyer 2003).

COP summary statistics describing AP, ML, and P COP trajectories were calculated based upon the recommendations of Prieto et al. (1996). The mean and range of the COP along each axis was first determined. For subsequent calculations, including the area enclosed by the COP per unit time (Prieto et al. 1996), the mean value was subtracted from each COP trajectory. For the ensuing analyses, the higher frequency components of the COP were attenuated using a fourth-order zero-phase low-pass Butterworth digital filter with a cutoff frequency of 7.5 Hz. The average power spectral density (PSD) of the COP along each axis was estimated using 2048-point Fast Fourier Transforms of 20-s windows overlapping by 0.5 s. The median power frequency was determined from each PSD with a resolution of 0.05 Hz. The mean velocity of the COP in the ML, AP, and P dimensions was calculated by dividing the total excursion of a trial by its duration. The magnitude of body COM accelerations was quantified using the root-mean-square (RMS) of the ground reaction shear forces. COP and shear force summary statistics from each trial were averaged to produce a single value for each combination of parameter, subject, and sensory condition.

For each balance task (bipedal-EC, bipedal-EO, unipedal-EO), stabilogram–diffusion parameters and summary statistics from each sensory condition (control, reduced sensitivity) were compared by 1-way repeated measures analysis of variance using the Statistica 5.5 software package (Statsoft Inc., Tulsa, OK, USA). In case of a significant effect of sensory condition, post hoc comparison was performed using Tukey's Honest Significant Difference test. Threshold for statistical significance was set at α=0.05.

Procedure 2: reduced sensation from the entire weight-bearing foot soles

In a follow-up experiment, sensation from the skin overlying the metatarsal heads, the lateral soles, and the heels (Fig. 1B) of the plantar soles of six subjects was reduced through multiple intradermal injections of an anesthetic solution (2% lidocaine HCl, 1:8 sodium bicarbonate, 1:200,000 epinephrine and 12 U/mg hyaluronidase, 30-gauge needles), administered by a board-certified anesthesiologist. Sodium bicarbonate served as a buffer to the acidic lidocaine, while epinephrine reduced the rate of washout and prolonged the effect of anesthesia. The use of hyaluronidase increased the area affected by each injection, thereby reducing the number of injections required. The toe pads remained untreated, since the use of epinephrine is precluded in the digits due to the risk of ischemia (Dollery 1991). Eight to fifteen 1-ml injections were required for each foot. Injections were performed very slowly to minimize subject discomfort, requiring a total of ~0.5 h for each foot. Pressure sensory threshold levels were assessed before and after balance testing in the manner described for procedure 1.

A Disk-Criminator (Neuroregen L.L.C., Bel Air, MD, USA) was used to determine two-point discrimination ability on the plantar skin overlying the third metatarsals and the heels of each foot before and after anesthesia. Using a two-alternative forced choice procedure, the closest prong-spacing that the subject correctly identified in at least two of three applications was designated the two-point discrimination threshold distance for that site.

Under control conditions and after anesthesia injection, changes in elastic skin compliance due to injection of anesthetic fluid were estimated using a modified spring-loaded dial position indicator. The indicator was fixed to a precision instrumentation stage and positioned to press the plunger against the sole of the right heel. Elastic compliance of the skin was approximated by a linear regression of the relationship $C_{skin} = (s-d)/k_{spring}d$, where s is the stage displacement, d is the position indicator displacement, and k_{spring} is the elastic stiffness of the modified position indicator. In order to minimize any difference in compliance between the two sensory conditions, subjects soaked their feet in warm tap water for 20 min prior to control testing. Changes in skin compliance were evaluated using a (within-subjects) Wilcoxon test.

Balance tests for the control condition were performed first, followed by reduced plantar sensation testing 1–7 days later. Under each of the two sensory conditions, subjects performed ten 35-s trials of bipedal standing with their eyes closed with 45 s of standing rest after each trial. Ground reaction forces were measured using two adjacent Kistler 9284 multi-component force platforms. Instructions to subjects were identical to those described for bipedal stance under procedure 1. Tape markings on the force platforms were used to ensure repeatable placement of the feet[2]. Postural sway was quantified according to procedures described for procedure 1. Results obtained before and after anesthesia were compared using Student's t-test for dependent samples with a threshold for statistical significance of α=0.05.

Results

Procedure 1: reduction of plantar forefoot sensitivity

Subjects reported no discomfort associated with alternating-pulse iontophoretic delivery of anesthesia, other than a mild tingling sensation under the dispersive pads during the first few minutes of the procedure. Prior to the application of anesthesia, the mean sensory threshold level (STL) was 3.97 a.u. (≈9 mN). Five minutes after the procedure, the average STL was increased to 5.56 a.u. (≈350 mN), while follow-up testing after the balance experiments showed an average STL of 5.00 a.u. (≈98 mN). All subjects achieved or exceeded the target STL of 5.07 a.u. (≈115 mN) after anesthesia. The procedure was therefore considered a successful simulation of a focal peripheral sensory neuropathy. Subjects gave correct answers for all trials of toe position sensation, confirming that intrinsic foot proprioception remained unaffected.

Statistically significant changes in balance parameters associated with plantar anesthesia are summarized in Table 1. An example of a stabilogram–diffusion function, calculated from real COP data in this study, is provided in Fig. 2 as a visual reference for some of the results provided below. The effect of forefoot anesthesia (Table 1, procedure 1) was most prominent during unipedal stance. The unipedal-EO results showed an increase in the velocity and median frequency of body sway. Mean COP velocity increased in all directions by approximately 10%, while the anteroposterior COP median frequency increased by an average of 0.09 Hz, or 36%. There was an increase in short-term planar diffusion coefficient, which is synonymous to a steeper slope of the line fitted to the short-term region of the linear representation of the stabilogram–diffusion function (cf. Fig. 2, top panel). This signifies an increase in stochastic activity over short time intervals (<~0.6 s in the example shown in Fig. 2).

During the bipedal-EC task, anesthesia increased mediolateral COP velocity and COP area by approxi-

[2] The subjects in procedure 2 also performed dynamic tests of balance that will be detailed in future publications. The unperturbed balance tests performed in procedure 2 were confined to a subset of those used in procedure 1 so that sufficient anesthesia could be maintained for both quasi-static and dynamic testing.

Table 1 Statistically significant effects of reduced plantar sensitivity. Data are changes in center-of-pressure (COP) parameters for unipedal and/or bipedal stance under procedure 1 (reduced plantar forefoot sensitivity) and procedure 2 (reduced sensation from the entire weight-bearing foot soles). *EO* eyes open, *EC* eyes closed, *P* plantar dimension, *AP* anteroposterior dimension, *ML* mediolateral dimension, *RMS* root-mean-square

Procedure	Task	COP parameter	Abbreviation	Change from control		
				Absolute	Percentage	*p*-Value
1	Unipedal-EO	P short-term diffusion	Ds_P	+21.1 mm^2/s	+18%	0.039
		AP median frequency	Mf_{AP}	+0.09 Hz	+36%	0.003
		ML velocity	Mv_{ML}	+2.2 mm/s	+9%	0.036
		AP velocity	Mv_{AP}	+1.9 mm/s	+11%	0.024
		P velocity	Mv_P	+3.2 mm/s	+9%	0.014
		AP shear force RMS	F_{AP}	+0.2 N	+11%	0.037
	Bipedal-EC	ML velocity	Mv_{ML}	+0.5 mm/s	+12%	0.004
		Mean COP area/s	Ma_P	+1.4 mm^2/s	+11%	0.049
		ML shear force RMS	F_{ML}	+0.05 N	+13%	0.022
2	Bipedal-EC	AP short-term scaling	Hs_{AP}	−0.03	–4%	0.038
		P velocity	Mv_P	+1.2 mm/s	+11%	0.051

mately 11%. Despite the anteroposterior asymmetry in anesthesia, there was no shift in the average COP position during any of the tasks. Shear forces were increased when the balance control system was taxed by closing the eyes (bipedal-EC, mediolateral) or unipedal stance (unipedal-EO, anteroposterior), indicating greater accelerations of the body COM. No statistically significant effects of forefoot anesthesia were found on bipedal stance when vision was available.

Procedure 2: reduced sensation from the entire weight-bearing foot soles

The intradermal injections of anesthesia caused only moderate discomfort and were well tolerated by all subjects. Unpleasant sensations were associated with the injection of fluid rather than the insertion of the 30-gauge needles. A Wilcoxon test of elastic skin compliance estimates confirmed that there was no significant change associated with the injection of fluid into the dermis (p>0.7, average change 9±28%). Likewise, there were no changes in subjects' ability to perceive movements of the toes. Subjects reported a sensation that their soles had dramatically increased in thickness but exhibited no obvious functional deficits after anesthesia.

Prior to anesthesia, average pressure STLs under the heels and first metatarsal heads were 4.14 and 4.15 a.u. (≈14 mN), respectively. Immediately following the anesthesia procedure, STLs were elevated to or above 6.65 a.u. (≈4 N, the stiffest filament) in 23 of 24 test sites (6 subjects × 4 test sites). After approximately 1 h of standing, 17 of 24 sites remained at or above 6.65 a.u. All tested sites were elevated above the target threshold of 5.07 a.u. (≈115 mN) for the duration of the balance experiments. Prior to anesthesia, the average two-point pressure discrimination threshold under the third metatarsal heads was 12.5 mm. After anesthesia, all thresholds exceeded the device maximum of 20 mm and remained so for the duration of balance testing.

Notable changes in balance parameters associated with whole-sole anesthesia are summarized in Table 1 (procedure 2). The 4% average decrease in short-term anteroposterior scaling exponent (Hs_{AP}) indicates reduced persistence in the COP trajectory over short time intervals (<~0.6s in the example shown in Fig. 2). A reduction in persistence is equal to a lesser slope of the line fitted to the short-term region of the log–log version of the stabilogram–diffusion function (Fig. 2, lower panel). The average COP velocity, however, was increased by 11% in the planar dimension.

Discussion

In contrast to previous research on postural feedback from foot afferents, the methods used in this study were designed to specifically target cutaneous foot-sole afferents for anesthesia. Anesthesia was introduced to the skin of the foot soles directly, preventing confounding effects on foot proprioception. The results indicate that reduced plantar forefoot sensation produced predominantly mediolateral balance deficits during bipedal stance, although only when vision was occluded. During unipedal stance with eyes open, the deficits were larger in magnitude and occurred in both the mediolateral and anteroposterior directions. In a follow-up experiment, loss of sensation from the entire weight-bearing foot soles produced changes in anteroposterior postural control during bipedal stance without vision. These findings demonstrate that feedback from other sensory mechanisms is insufficient to fully compensate for reduced plantar sensitivity when the balance control system is challenged by unipedal stance or the loss of visual input.

A consistent finding in both anesthesia experiments was an increase in COP velocity when the posture control system was challenged by unipedal stance or deprived of visual cues. COP ranges were not increased, nor were changes seen in the transition displacement. In general, then, the magnitude of postural sway displacement was not affected by foot-sole anesthesia. Increases in shear force RMS, however, indicate that the magnitude of body COM accelerations did increase after forefoot anesthesia. Our results are generally consistent with previous studies of somatosensory contributions to posture. For example, anteroposterior COP velocity was increased with eyes closed after both feet were anesthetized by cold-water immersion below the ankles (Magnusson et al. 1990b). No increase in body sway amplitude during unperturbed stance was found after bilateral hypoxic ischemia of the feet and ankles (Horak et al. 1990). Likewise, COP range during unipedal stance was not increased after anesthetic block of the feet and ankles (Konradsen et al. 1993). Reconstructive plantar surgery causing reduced sensation from a portion of one foot sole increased COP velocity (Hämäläinen et al. 1992). As in the present study, increased shear force production has been seen in peripheral sensory neuropathy patients during anteroposterior sensory conflicts (Simmons et al. 1997). In contrast to the present study, however, Asai et al. (1990) reported an average posterior shift in the COP with no increase in COP velocity after chilling both soles with cryogenic air. Whole-sole cooling with cold air also increased COP area, whereas we found this effect only when anesthesia was confined to the forefoot. Since the five subjects reported by Asai and colleagues only performed a single 20-s trial under each sensory condition, these inconsistencies with respect to the present study may simply be related to the significant inter-trial variability seen in many postural parameters (Meyer 2003).

It is interesting to compare the effects of forefoot and whole-sole anesthesia. Forefoot anesthesia produced mostly mediolateral postural effects, increasing the mediolateral COP velocity during both bipedal-EC and unipedal-EO tasks. Whole-sole anesthesia, however, appeared to produce an anteroposterior effect, decreasing persistence over short time intervals as well as increasing COP velocity[3]. These differences could simply be related to the difference in depth of anesthesia produced by the two procedures. It is likely, however, that the loss of sensation from the plantar heels was partly responsible for these different effects. Perry et al. (2000) conducted a study of dynamic balance after reducing plantar sensitivity with cold water. They concluded that during forward support surface translations, when body weight is rapidly transferred to the heels, heel sensation is particularly important for generating successful stepping responses. Balance-related information from the cutaneous forefoot is probably supplemented by proprioception from the toes and intrinsic foot muscles. In contrast, relatively little proprioceptive information regarding heel contact is available. It is reasonable then to hypothesize that plantar sensation from the heels may influence the control of anteroposterior balance, while plantar forefoot sensation may be more important for mediolateral control.

Collins and De Luca (1993) hypothesized that the existence of different behavior over short and long time scales within COP trajectories may be indicative of open- and closed-loop control mechanisms, respectively. In their conceptual model, these two control systems operate concurrently. Open-loop control operates independent of sensory feedback, while the closed-loop behavior manifests the feedback control of posture. The anesthesia procedures used in the current study increased the threshold for the perception of plantar pressures. Presumably, there was also an increase in the threshold for the activation of any plantar-pressure triggered postural corrections. Rather than affecting transition point coordinates, however, plantar anesthesia produced only moderate increases in COP velocity and short-term diffusion coefficients. In terms of the Collins and De Luca model, our results generally reflect changes in the open-loop control of posture, with little or no effect on the closed-loop control mechanisms. Consequently, foot-sole pressure sensation may not be heavily involved in the feedback control of posture. Rather, plantar feedback may be used to determine the appropriate level of background muscle activity based upon characteristics of the support surface. In contrast to the open-loop/closed-loop control hypothesis, Peterka (2000) proposed an alternate explanation for the stereotypical shape of the stabilogram–diffusion function. He demonstrated that a simple linear proportional-integral-derivative (PID) feedback model without the existence of sensory detection thresholds produced physiologically plausible stabilogram–diffusion functions. From his results, it does not appear that modulation of any single parameter in the model can replicate the results of the present study. For instance, if we assume that Peterka's proportional element is analogous to ankle stiffness, our results are inconsistent with an increase in stiffness such as that seen when subjects perceive a postural threat (Carpenter et al. 2001).

Morasso et al. (1999) have suggested a feedback model similar to that of Peterka (2000) in which plantar pressure sensation is used to derive an estimate for the position of the body COM. Like Peterka's model, that of Morasso and colleagues is based upon the common assertion that the major task of the posture control system is to maintain the position of the body COM over the base of support. Morasso et al. (1999) demonstrated mathematically that an estimate for the position of the COM could be derived from the shear force under the feet and the location of the COP. If COM position is a variable controlled by the posture control system, reliance on plantar sensation for COM feedback after foot-sole anesthesia would likely result in a dramatic decrease in stability. Our experimental evidence, however, demonstrates that even complete plantar anesthesia produces only moderate changes in unperturbed balance. We can therefore conclude that plantar afferents are not the primary source of feedback regarding the position of the body COM.

We found that reduced cutaneous foot-sole sensation resulted in increased average COP velocity when subject's eyes were closed. The ~11% increases in COP velocity seen in the present study, however, were considerably smaller than the ~40% (Boucher et al. 1995) to 60% (Simoneau et al. 1994) increases attributed to diabetic

[3] Statistically significant increases in COP velocity after whole-sole anesthesia were seen only in the planar dimension. A notable increasing trend (p=0.066) in anteroposterior COP velocity suggests that the increase in planar COP velocity may be primarily due to increases in AP COP velocity.

peripheral neuropathy under similar conditions. Peripheral neuropathy patients also exhibited increases in COP range of 27–45% (Cavanagh et al. 1993; Boucher et al. 1995) during unperturbed stance, whereas we found that reduced plantar sensitivity produced no such change. Plantar cutaneous anesthesia had no effect on standing balance when vision was available, whereas the effects of peripheral neuropathy were similar with and without visual input. It appears that even complete loss of sensation from the weight-bearing cutaneous foot sole is insufficient to replicate much of the balance dysfunction associated with peripheral neuropathy. This finding could indicate that cutaneous sensory losses are less important than proprioceptive deficits in producing the balance complications associated with diabetic peripheral neuropathy. It is likely, however, that the cumulative effect of multiple sensory losses is highly non-linear. An example of this is seen in the data on unperturbed stance in peripheral neuropathy patients presented by Simoneau et al. (1994). Considering healthy subjects with head straight and eyes open as controls, tilting the head back to reduce the effectiveness of vestibular input induced virtually no effect on COP velocity. However, tilting the head added ~10% (relative to control) to the effect of closing the eyes, and an additional ~100% to the combined effects of eye closure and peripheral neuropathy. Similarly, the forefoot-sole anesthesia used in the present study had no effect on balance when vision was available, but approximately doubled the effects of closing the eyes on mediolateral COP velocity and shear force RMS. It seems that the balance instability associated with the loss of multiple sensory modalities may be far greater than sum of the effects of individual sensory losses. We can therefore expect that the effects of reduced plantar sensation on standing balance will be amplified by the concomitant proprioceptive deficits that typically follow from diabetic peripheral neuropathy.

Conclusion

Considerable theoretical and experimental evidence exists to suggest that feedback from plantar cutaneous afferents may be important for the maintenance of upright stance. Previous balance studies, however, have largely failed to isolate the role of plantar cutaneous afferents from other foot and ankle somatosensory systems. In contrast, the present study targeted the cutaneous foot soles for anesthesia, eliminating confounding effects on foot and ankle proprioceptors as well as the intrinsic foot musculature. Reduced plantar sensation had no effect on bipedal balance when vision was available. When the postural control system was challenged by unipedal stance or closing of the eyes, the loss of plantar sensation caused an increase in the velocity of postural sway. Our results indicate that feedback from plantar cutaneous afferents is important for the maintenance of normal balance when vision is not available. The importance of plantar sensation is expected to increase with the concomitant proprioceptive deficits commonly seen in peripheral neuropathy patients.

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