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RESEARCH ARTICLE

Postural control during kneeling

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Abstract Postural control was studied when the subject was kneeling with erect trunk in a quiet posture and compared to that obtained during quiet standing. The analysis was based on the center of pressure motion in the sagittal plane (CPx), both in the time and in the frequency domains. One could assume that postural control during kneeling would be poorer than in standing because it is a less natural posture. This could cause a higher CPx variability. The power spectral density (PSD) of the CPx obtained from the experimental data in the kneeling position (KN) showed a significant *decrease* at frequencies below 0.3 Hz compared to upright (UP) ($P < 0.01$), which indicates less sway in KN. Conversely, there was an *increase* in fast postural oscillations (above 0.7 Hz) during KN compared to UP ($P < 0.05$). The root mean square (RMS) of the CPx was higher for UP ($P < 0.01$) while the mean velocity (MV) was higher during KN ($P < 0.05$). Lack of vision had a significant effect on the PSD and the parameters estimated from the CPx in both positions. We also sought to verify whether the changes in the PSD of the CPx found between the UP and KN positions were exclusively due to biomechanical factors (e.g., lowered center of gravity), or also reflected changes in the neural processes involved in the control of balance. To reach this goal, besides the experimental approach, a simple feedback model (a PID neural system, with added neural noise and controlling an inverted pendulum) was used to simulate postural sway in both conditions (in KN the pendulum was shortened, the mass and the moment of inertia were decreased). A parameter optimization method was used to fit the CPx power spectrum given by the model to that obtained experimentally. The results indicated that the changed anthropometric parameters in KN would indeed cause a large decrease in the power spectrum at low frequencies. However, the model fitting also showed that there were considerable changes also in the neural subsystem when the subject went from standing to kneeling. There was a lowering of the proportional and derivative gains and an increase in the neural noise power. Additional increases in the neural noise power were found also when the subject closed his eyes.

Keywords Human · Power spectral density · Center of pressure · Model · Neural noise · Inverted pendulum

Introduction

Kneeling is a position employed in a variety of work environments and situations. Although the ergonomic aspects of subjects during kneeling have been evaluated (Gallagher and Unger 1990; Gallagher 2005), no study has been found that focused specifically on the problem of postural control. As kneeling is an unnatural position one could think that balance control would be poorer than that found during quiet upright stance.

From an applications point of view, the study of postural control during kneeling may be relevant, for example, in the evaluation of different demanding tasks (e.g., lifting a weight) in a variety of work environments (Gallagher and Unger 1990; Gallagher 2005; Splittstoesser et al. 2007). In addition, in subjects with some motor

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disorders, kneeling is a transitional position used to attain the upright posture, e.g., after a fall. An improved knowledge of the control of stability during kneeling could be useful for physiotherapists, as they employ different technical procedures based on this position. For example, patients could be trained in the kneeling posture to establish and reinforce appropriate synergies to achieve the task of getting up (Vander Linden and Wilhelm 1991). Physiotherapy approaches during kneeling may also focus on the facilitation of hip muscle tone (e.g., gluteus medius) in order to stabilize postural alignment in the frontal plane during gait.

The general question posed in this research is how does the postural control system change its performance when the subject is in a kneeling position. Will there be a larger variability of the position of the center of pressure (CP)? Will the central nervous system (CNS) employ the same dynamics as during standing or will the CNS change its dynamic characteristics to fit the actual postural challenge? An increased variability in the CP position could be due to a poorer proprioceptive feedback or a poorer tuning of the CNS to control such a position, among other causes of neural origin. Alternatively, the CP variability could decrease because in the kneeling position the center of mass is situated closer to the ground, which is inherently easier to stabilize.

A final topic in this study was to describe the effect of vision on CP variability in the two postures and check if its influence on the control of balance changes from one condition to another. The effect of lack of visual input has been extensively investigated in the upright stance in young, aged and motor-impaired populations (Perrin et al. 1997; Lafond et al. 2004; Cornilleau-Pérés et al. 2005; Vaillant et al. 2008). A significant effect of vision on CP parameters during upright stance has been reported mainly for the mean velocity (Riach and Starkes 1994; Raymakers et al. 2005; Baccini et al. 2007; Slobounov et al. 2008), even though the effects were found to be distributed over the whole power spectrum (Mezzarane and Kohn 2007).

The approaches to answer the questions posed above were both experimental and through modeling. In the former, the main tools for analysis were the power spectrum of the CP in the antero-posterior direction (CPx) and a few time domain parameters. In the latter, a simple inverted pendulum postural control system was adapted from the literature (Peterka 2000; Maurer and Peterka 2005) and its theoretical CPx power spectrum was fitted to the experimentally obtained spectrum either from standing or kneeling subjects. The fitting was achieved by an optimization method that varied a few chosen model parameters in order to minimize the error between the experimental and theoretical power spectra.

Methods

Subjects

The experiments were carried out on 12 healthy subjects (7 males and 5 females) aged 32.4 ± 7.1 years, with 1.73 ± 0.1 m height and weighing 68.2 ± 12.7 kg. All the participants gave their informed consent according to the local ethics committee.

Procedures

Subjects stood over the center of a force platform (AMTI, OR6-7-1000) in four different conditions: (1) Quiet upright stance with eyes open (UP-EO); (2) Quiet upright stance with eyes closed (UP-EC); (3) Kneeling position with eyes open (KN-EO); (4) Kneeling position with eyes closed (KN-EC).

In the upright (UP) and kneeling (KN) positions the feet and knees were spaced at shoulder width. The force platform was embedded in an elevated floor and the posterior part of the floor (relative to the force platform) was withdrawn in KN condition to maintain the leg (and the ankle joint) in a straight angle (Fig. 1a, right panel). A thin foam of low density was used to avoid discomfort at the knee due to its contact with the hard surface of the platform.

All subjects remained over the platform in each condition for 70 s. The first 10 s of acquisition were discarded. In EO conditions the subjects stared at a target at the eye level located at the distance of 3.6 m. In both positions, the subjects were instructed to stay quietly and to minimize movements of their hips. Two trials in each condition were performed in all subjects (with an interval of 2 min between them).

In a different day, each subject stayed over a pressure/force-sensing floor mat (MatScan 4.21, Teckscan) for 60 s in order to estimate the length of the base of support in the antero-posterior direction in both positions (KN and UP) (see Fig. 1b).

Signal processing

The forces and moments relative to the 3 orthogonal axes (X, Y and Z; see the axes in Fig. 1b) provided by the force platform were used to evaluate the center of pressure in the antero-posterior direction (CPx). All signals were fed into a PC-based DataWave data acquisition and processing system that sampled each channel at a frequency of 100 Hz. The signal acquisition system generated data files in ASCII format that were processed in MATLAB (6.5, MathWorks) environment.

The root mean square (RMS), the mean velocity (MV), and the power spectral density (PSD) of the CPx signal

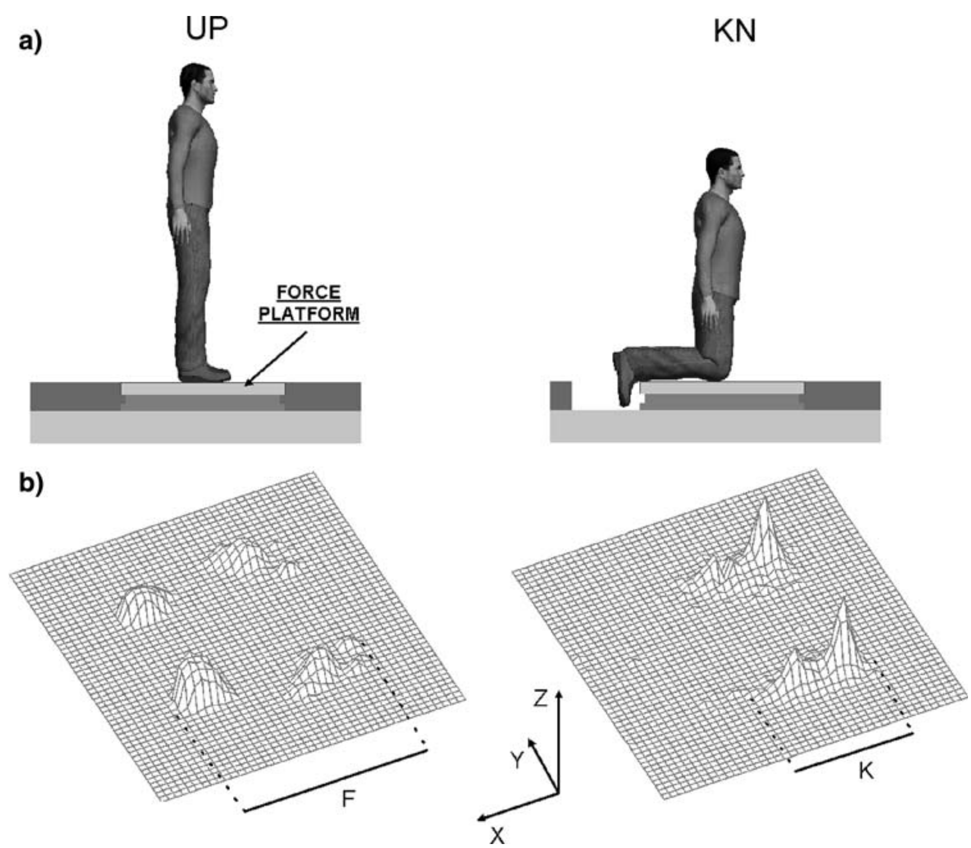

Fig. 1 **a** Subject over the force platform in both positions: upright (UP) and kneeling (KN). **b** Pressure distribution under the feet and knees to evaluate the length of the base of support F and K, respectively. Calibration $Z = 100$ kPa; Y and $X = 10$ cm

were evaluated for each trial in each condition. The RMS and MV were computed as follows:

$$\mathrm{RMS} = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(\mathrm{CPx}_i)^2}$$

$$\mathrm{MV} = \frac{fs}{N}\sum_{i=2}^{N}|\mathrm{CPx}_i - \mathrm{CPx}_{i-1}|$$

where N is the number of samples of the CPx signal and fs is the sampling frequency (100 Hz).

The averaged PSD was evaluated from all trials of the 12 subjects ($n = 24$). The PSD of the CPx from each trial was estimated using the Welch method with 2,000 samples per periodogram, resulting in a spectral resolution of 0.05 Hz. A Hann data-window was used with subtraction of the best linear regression and an overlap of 1,000 samples.

The area under the PSD was evaluated for each trial at the following frequency ranges: "low" (0.05–0.25 Hz), "medium" (0.3–0.7 Hz) and "high" (0.75–2.0 Hz) (see Fig. 4a). The criteria used to establish the frequency ranges were based on previous findings from the literature and the profile of the spectra. The upper limit of the low frequency range (0.25 Hz) was chosen based on the fact that the spectra of both CPx and center of gravity displacements are very similar at least up to 0.2 Hz (Gage et al. 2004). The choice of the upper limit of the medium frequency range (0.7 Hz) was based on the profile of the spectrum itself, i.e., there was an inversion in the pattern of power distribution between UP-EO and KN-EO conditions at 0.7 Hz (indicated by an arrow in Fig. 5c, left panel). The upper limit of the high-frequency range was chosen as 2.0 Hz because 99% of the spectral power was found to be below this frequency in UP-EO condition.

Signals from the pressure-sensing floor mat were acquired with a sampling frequency of 10 Hz (10 frames per s). Each frame was used to compute an estimate of the distance between the most anterior and the most posterior points in which the pressure remained above 0.002% of that associated with the total weight for each subject (Fig. 1b). Pressure values below this chosen critical level were due to the presence of the foam.

In order to quantify the differences in the body oscillations relative to the respective length of the base of support between UP and KN, the CPx was low-pass filtered with a cutoff frequency of 0.3 Hz (only in the EO condition). The RMS values were then normalized by the length of the corresponding base of support (F or K from Fig. 1b) for each subject.

Statistical analysis

The effects of position and vision were analyzed using a two-factor ANOVA of repeated measures. A Bonferroni post-test was applied to detect differences between the levels. For all variables tested the differences between conditions were considered significant at $P < 0.05$. All the analyses were performed using the statistical package SPSS.

System modeling

The model used as a research aid in the present study is based on Maurer and Peterka (Peterka 2000; Maurer and Peterka 2005) for the postural control system of a standing human. Its block diagram may be seen in Fig. 2. It shows a block G_n which corresponds to the neural subsystem of the postural control system, a feedback loop delay τ_d, an inverted pendulum description of the standing human, and the transformation of postural angle to the CPx. The reference level was set at 0 without loss of generality. The model was implemented in Simulink (MathWorks).

The block G_n is a transfer function that has the form of a proportional, integrative and derivative system (PID) and represents the action of the nervous system as a controller as well as the properties of muscle and sensory receptors. Mathematically, G_n is defined as: $G_n = sK_d + K_p + K_I/s$, where s is the complex variable used in Laplace transforms, K_d is the derivative gain (Nms/deg), K_p is the proportional gain (Nm/deg) and K_I (Nm/s.deg) is the integrative gain. The human being during quiet stance is approximated by an inverted pendulum, with J_b the moment of inertia, m the mass, d the height of the center of mass and g the acceleration due to gravity. Neural (or torque) noise is represented by filtering a band-limited white noise process by an appropriate lowpass filter defined by the parameters K_n and τ_n. An increase in neural noise is associated with an increase in the gain K_n. The angle θ (radians) of the subject with respect to vertical is sensed by receptors and is fed back to the nervous system. The term G_r in this model is simply a constant value necessary to transform radians into degrees to follow the units of K_d, K_p and K_I used here. The block in the feedback loop indicated by τ_d indicates the total delay in the postural control loop, which includes the afferent and efferent pathways plus processing time in the central nervous system. The output of interest is the CPx, whose power spectrum will be indicated by $S_{cc}(\omega)$

The system in Fig. 2 is linear, since the angle θ was assumed to be small enough so that some known nonlinearities can be neglected. The power spectrum of the CPx may be obtained from the block diagram in Fig. 2 as a function of the system components, since the input is a (band-limited) white noise with unit variance. Skipping the mathematics, the expression of the CPx power spectrum $S_{cc}(\omega)$ as a function of angular frequency ω in radians/s, is given by

$$S_{cc}(\omega) = |G_{TOT}(\omega)|^2 \cdot \frac{K_n^2}{\omega^2\tau_n^2 + 1} \quad (1)$$

where

$$G_{TOT}(\omega) = \frac{(d + \omega^2 \cdot J_b/mg)}{-\omega^2 J_b - mgd + G_r\left(K_p + j\omega K_d + K_I/j\omega\right)\exp(-j\omega\tau_d)} \quad (2)$$

The following parameter values were adopted for a standing subject, based on (Peterka 2000; Maurer and Peterka 2005): $\tau_n = 100$ s, $J_b = 66$ kg.m^2, $m = 76$ kg, $g = 9.8$ m/s^2, $d = 0.87$ m, $K_I = 0.60$ Nms^{-1}deg^{-1}, $\tau_d = 0.1$ s.

The purpose of using the model above is to try to answer the question if there is a change in K_n (related to the neural noise power) and the model parameters K_p and K_d, when the subjects are in a kneeling position as compared to

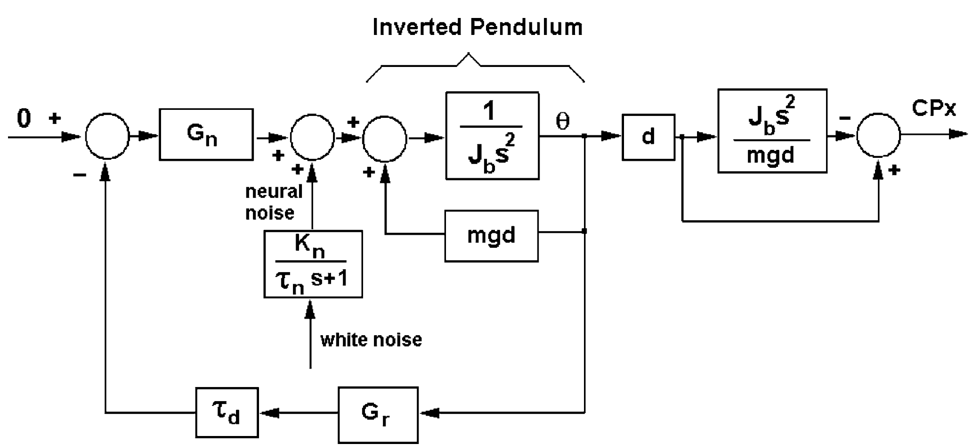

Fig. 2 Block diagram of the model of postural control. The output of the inverted pendulum, which is the angle with respect to vertical, is fedback to the central nervous system controller whose dynamics are embedded in G_n. The output of the system is the center of pressure in the antero-posterior direction, CPx

standing. Recently, Maurer and Peterka (Maurer and Peterka 2005) and van der Kooij et al (van der Kooij et al. 2005) have presented approaches for system identification applied to postural control models, which are very useful to estimate model parameters from the CPx time course.

In the present work we chose an unconstrained nonlinear optimization (Nelder and Mead) method to estimate the parameters K_p, K_d and K_n from the CPx power spectrum. The experimentally obtained average power spectrum for the subjects in one of the two conditions (UP-EO or KN-EO) was sampled at every 0.1 Hz from 0.1 to 2 Hz and at every 0.4 Hz from 2 to 7 Hz. The value of 7 Hz was chosen ad hoc (instrumentation or background noise was still negligible but the frequency was high enough from a mathematical sense, as will be seen in what follows) and is not critical. The different frequency resolutions for the optimization method were found to be required because when a 0.1-Hz resolution was used for the 2 to 7 Hz range, the optimization method gave too much importance to fitting the long and slowly decaying tail of the power spectrum (this improved the estimate of K_n but worsened that of K_p and K_d). The function *fminsearch* from MATLAB was used to estimate the values of the three parameters by minimizing a normalized absolute error E between the experimental power spectrum $S_{\mathrm{ccexp}}(\omega)$ and that given by formulas (1) and (2) at the selected frequencies ω_i, $i = 1,\ldots, N$:

$$E = \sum_{i=1}^{N} \left| \frac{S_{\mathrm{cc}}(\omega_i) - S_{\mathrm{ccexp}}(\omega_i)}{S_{\mathrm{ccexp}}(\omega_i)} \right|$$

As an alternative method, the value of K_n was found to be reasonably well estimated, *under the typical values of the model parameters given in the literature*, by computing the theoretical power spectrum, given by (1) and (2), at $\omega_{\mathrm{high}} = 2.\pi.7$ rad/s. The value of $S_{\mathrm{cc}}(\omega)$ at $\omega = \omega_{\mathrm{high}}$ is *approximated* by

$$S_{\mathrm{cc}}(\omega_{\mathrm{high}}) \approx \frac{K_n^2}{\omega_{\mathrm{high}}^2 m^2 g^2 \tau_n^2} \tag{3}$$

Therefore, an approximate estimate of K_n can be obtained from the experimental power spectrum $S_{\mathrm{ccexp}}(\omega_{\mathrm{high}})$ as

$$K_n \approx \sqrt{S_{\mathrm{ccexp}}(\omega_{\mathrm{high}})}\,\omega_{\mathrm{high}} m g \tau_n \tag{4}$$

The experimental spectral value at $\omega_{\mathrm{high}} = 2\pi\cdot 7$ rad/s in the formula above was actually computed as an average of the spectral values from $2\pi\cdot 6.75$ to $2\pi\cdot 7.25$ rad/s to increase spectral smoothness. Simulation runs with the model of Fig. 2 indicated that formula (4) provided an estimate with an error below 10% the nominal value.

Another formula may be derived that is useful to estimate the value of K_p. The theoretical value of the power spectrum, given by (1) and (2), is computed at $\omega_{low} = 2.\pi.0.1$ rad/s. The value of $S_{\mathrm{cc}}(\omega)$ at $\omega = \omega_{\mathrm{low}}$ and employing typical values for the model is approximated by

$$S_{\mathrm{cc}}(\omega_{\mathrm{low}}) \approx \frac{d^2 K_n^2}{\omega_{\mathrm{low}}^2 \tau_n^2 (G_r K_p - mgd)^2} \tag{5}$$

Formula (5), to be used after having estimated K_n from formula (4), yields a second degree equation in K_p. Simulations of the model in Fig. 2 showed that the largest solution for K_p obtained from the equation gives the correct answer. Mathematically, the value for K_p is, therefore, found as the largest root of the polynomial:

$$K_p^2 - \frac{2mgd}{G_r} K_p + \frac{(mgd)^2}{G_r^2} - \frac{(d \cdot K_n)^2}{S_{\mathrm{cc}}(\omega_{\mathrm{low}})(G_r \omega_{\mathrm{low}} \tau_n)^2} \tag{6}$$

These approximate formulas (4) and (6) were used in this work as an aid to check if the optimization method converged appropriately. All reported results obtained by the optimization method agreed well with the approximate values predicted by the formulas.

The anthropometric values for a subject during KN were computed based on data presented in Winter (1990). The body mass, height of the center of mass and moment of inertia of a kneeling subject resulted $m_{\mathrm{KN}} = 68.7$ kg, $d_{\mathrm{KN}} = 0.51$ m and $J_{\mathrm{KN}} = 21.5$ kg.m^2. The value of the loop delay when the subject is in position KN was estimated to be 86 ms (as compared with 100 ms for UP), based on published data on the latency differences between the Achilles tendon and patellar reflexes (Frijns et al. 1997). This change in the value of delay when kneeling was based solely on the proprioceptive feedback. However, vision and vestibular sensory inputs, if modeled separately, would be associated with only a slightly higher delay in KN, about 93 ms, (an 8 % difference over the 86 ms value) to account for the shorter path to the efferent muscles.

Results

This section will present first the experimental results, followed by the results obtained from the modeling.

Experimental results

The raw data from a single subject (Fig. 3a) illustrate some of the differences found in the CPx signals obtained from all the subjects in both positions. There were less low-frequency CPx oscillations in KN as compared to UP and larger relative amplitude of high-frequency CPx oscillations in KN. The data from this subject suggests less postural sway in KN as implied by the smaller PSD values of the corresponding CPx at low frequencies (Fig. 3b).

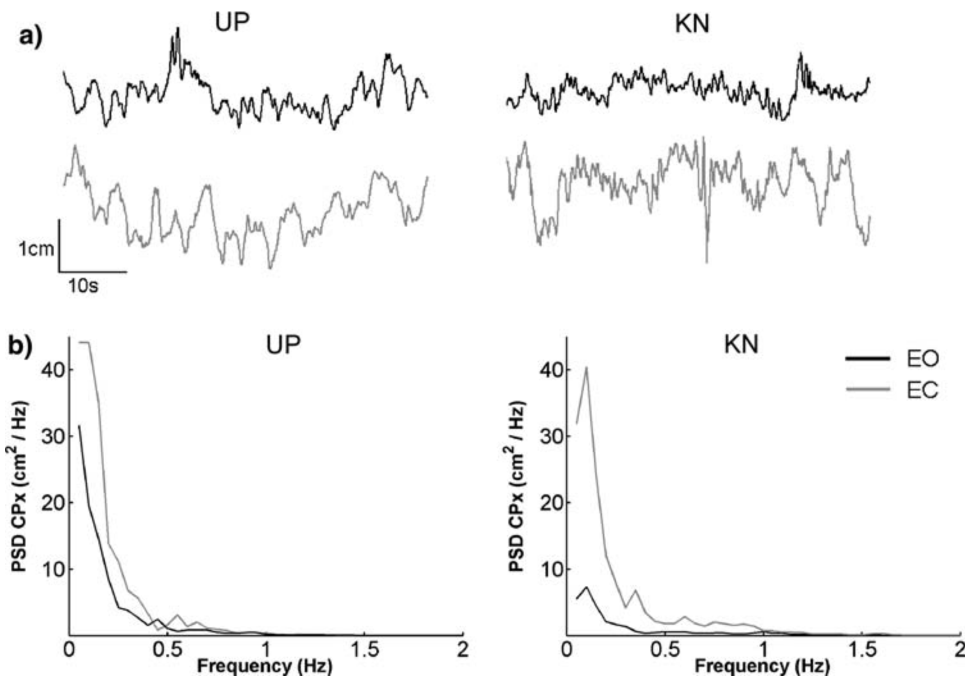

Fig. 3 Raw data from one subject. **a** CPx with eyes open (*black lines*) and closed (*gray lines*) obtained in both positions, UP (*left panel*) and KN (*right panel*). Upward deflection means forward displacement of the CPx. **b** Corresponding spectra estimated from the two trials in UP (*left panel*) and KN (*right panel*). Note the decreased power at low frequencies in KN-EO condition

The overall mean spectra (Fig. 4a) obtained in the upright and kneeling positions suggest less postural sway in KN due to the lower PSD values at low frequencies. A first quantitative corroboration came from the RMS value of the CPx, which showed larger values in UP when compared to KN ($F_{(1,11)} = 16.91$; $P = 0.002$). In contrast, the parameter MV increased in KN as compared to UP ($F_{(1,11)} = 6.96$; $P = 0.023$) (Fig. 4b, c). A second quantitative corroboration of less postural sway in KN came from the higher area under the PSD in the low-frequency range (below 0.3 Hz) for UP ($F_{(1,11)} = 12.36$; $P = 0.005$) (Fig. 5a). On the other hand, for the high-frequency range (above 0.7 Hz) the pattern was reversed, increased area under PSD in KN ($F_{(1,11)} = 5.05$; $P = 0.046$) (Fig. 5c). The latter result is consistent with that obtained for the MV parameter (compare Figs. 4c to 5c). Finally, the effect of position in the medium frequency range was not significant ($F_{(1,11)} = 0.514$; $P = 0.488$) (Fig. 5b).

Due to the power spectrum distribution, the RMS value within the interval of low frequencies was 86% of the total RMS, which may justify the consistency between RMS and low frequency differences between KN and UP.

The PSD values for UP-EC below 0.4 Hz were higher than those for KN-EC (Fig. 5b, left panel), and at frequencies above 0.4 Hz the opposite was observed, causing similar areas under the respective PSD in the medium frequency range (Fig. 5b, right panel). ANOVA results revealed that lack of the visual input significantly increased CPx variability, as measured by the time domain variable

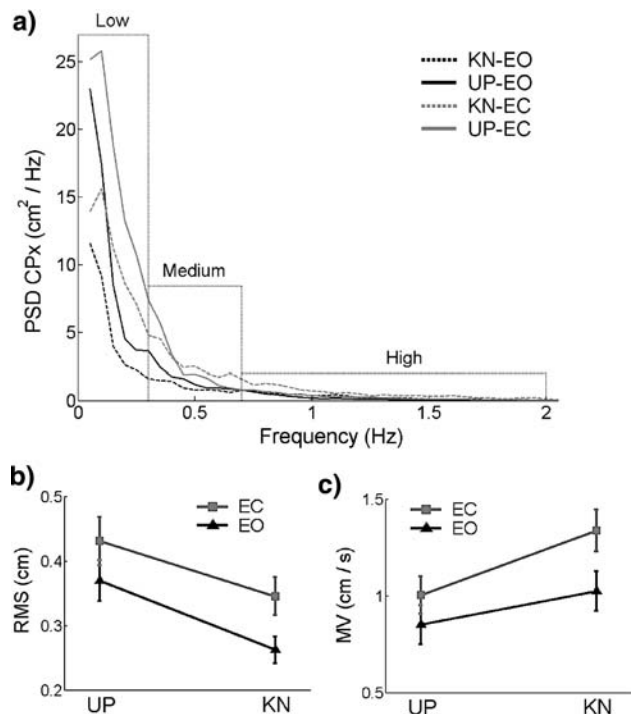

Fig. 4 **a** Average PSDs estimated from all subjects, in all conditions tested. The frequency ranges are delimited by *dotted lines*. **b** Mean values of the RMS of the CPx evaluated for both positions during open eyes (*black triangles*) and closed eyes (*gray squares*). Vertical lines indicate the SEM. **c** The same as in **b** for the MV. The effect of position over both measures (RMS and MV) was statistically significant. There was significant interaction between vision and position for the MV ($P < 0.05$)

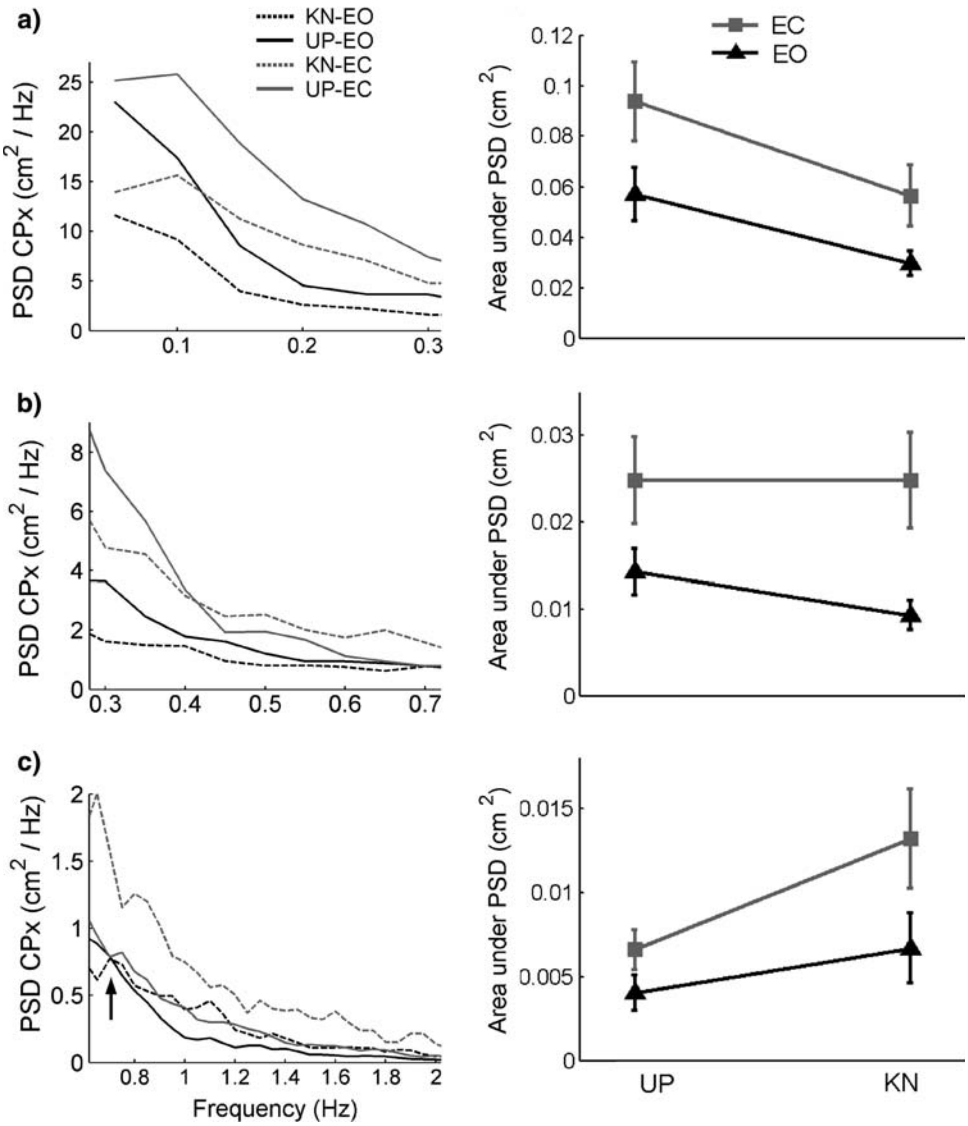

Fig. 5 Average PSDs redrawn from Fig. 4a for different frequency ranges (*left panels*) and associated areas (*right panels*): **a** PSD and area values under the "low" frequency range (0.05–0.25 Hz) evaluated for all conditions tested, **b** and **c** The same as in **a** for the "medium" (0.3–0.7 Hz) and "high" (0.75–2 Hz) frequency ranges. *Vertical lines* in the *right panels* indicate the SEM. The effect of position over the area values was statistically significant, except for the medium frequency range. Absence of vision affected the values of area under all frequency ranges. There was significant interaction between factors vision and position in the high frequency range ($P < 0.05$). The *arrow* in **c** indicates the crossing of the PSDs of UP-EO and KN-EO conditions (see "Methods")

RMS ($F_{(1,11)} = 8.56$; $P = 0.014$) and MV ($F_{(1,11)} = 31.75$; $P < 0.001$), as well as by frequency domain measurements, such as the spectral power in the low ($F_{(1,11)} = 7.30$; $P = 0.021$), medium ($F_{(1,11)} = 25.2$; $P < 0.001$) and high-frequency ranges ($F_{(1,11)} = 23.3$; $P = 0.001$) (Figs. 4, 5). Looking at the spectra, a first impression is that the effect of vision is similar in both postures UP and KN (Fig. 4a). However, there was a significant interaction between factors position and vision for the parameter MV ($F_{(1,11)} = 5.08$; $P = 0.046$) and the area under the high-frequency range ($F_{(1,11)} = 5.80$; $P = 0.035$). These results indicate that the effect of vision on the fast body oscillations was more emphasized in KN.

The apparent similarity of the CPx spectral profiles for the UP and KN postures may be better examined by multiplying the KN-EO spectrum by a constant such that the first spectral value coincides with that for the UP-EO

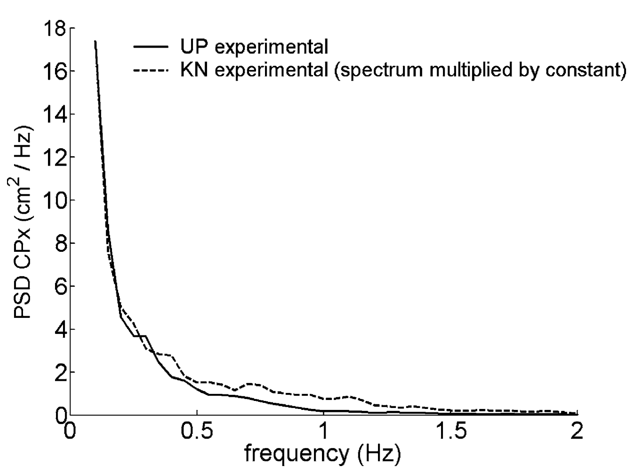

Fig. 6 Experimental power spectra of the CPx for the UP-EO and KN-EO conditions, with the KN spectrum multiplied by a constant so that the two spectra are equal at frequency equal to 0.1 Hz

condition. The result, seen in Fig. 6, indicates that there is some similarity in spectral shape but for the KN-EO condition the PSD is higher for frequencies above about 0.5 Hz.

The pressure distribution measured with the pressure-sensing floor mat system showed that the length of the base of support in KN (knee) was 52.3 ± 0.06% lower than in UP (feet) (see K and F lengths in Fig. 1b). However, the excursion of the slower component of the CPx (measured after filtering the CPx by a low-pass filter) relative to the total length of the base of support was similar for both conditions: 1.7 ± 0.6% and 2.1 ± 0.7% for UP and KN, respectively (no statistical significance; paired *t*-test: $P > 0.05$).

There were no differences in the values of the vertical force (Fz) for both positions (68.18 ± 12.74 kg and 68.12 ± 12.70 kg for UP and KN, respectively) (paired *t*-test: $P > 0.05$). This indicates that the contact of the toes with the floor (below the surface of the force platform) during kneeling, as observed in a few subjects, did not provide a mechanical support that could result in an increase of the base of support.

Model results

An initial hypothesis for the smaller sway found during kneeling was that it was solely due to the change in biomechanics. A lowered center of gravity, a smaller moment of inertia and a smaller loop delay all contribute to increased stability. In the following, the data refer to the open eyes condition, unless otherwise stated.

Indeed, formulas (1) and (2) used with the new values of biomechanical parameters showed a considerable decrease in sway, as indicated by the substantial decrease in PSD at low frequencies (dashed lines in Fig. 7a). However, from Fig. 7 it is also clear that the decrease in power spectral values was much larger than what was found in real life (dashed lines compared with continuous lines). Also, the changes in model parameters due only to the biomechanics caused a small resonance peak around 2 Hz (Fig. 7b, dashed line), which was not found in the experiments (Fig. 7b, continuous line). Therefore, it is clear that there were also changes in the parameter values of nervous system controller besides those associated with the altered biomechanics in KN.

The next logical step was to permit changes in the parameters associated with the nervous system, K_p, K_d, and K_n. Parameter K_I was left fixed because its influence in the dynamics of the postural control model is quite small (Maurer and Peterka 2005). The optimization search for the parameter values was applied first to the standing CPx average power spectrum of all the subjects, giving the estimated values shown in the second column of Table 1.

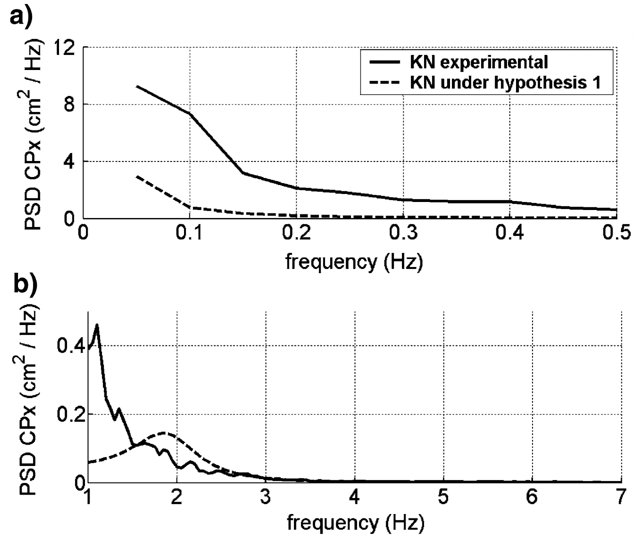

Fig. 7 Power spectra at two different frequency ranges, **a** and **b**, for KN-EO condition: experimental (*continuous line*) and obtained from the model (*dashed line*). The latter was computed from formulas (1) and (2) using the biomechanical parameters for the kneeling position and assuming the neural system remains with the same parameter values as in UP

Formulas (4) and (6) were used to provide parameter estimates that could increase the confidence on the appropriate convergence of the optimization method. When formula (4) was applied to the experimental average spectrum of the standing subjects, the result was $K_n =$ 718.1 Nm, which is near the result obtained by the optimization method (Table 1: $K_n = 728.7$ Nm). The largest root of (6), using the average spectrum of the standing subjects, resulted $K_p = 15.47$ Nm/deg, which is not far from that obtained by the optimization method (Table 1: $K_p = 16.8$ Nm/deg).

For the kneeling condition, the optimization method yielded the values for K_p, K_d, and K_n shown in the fourth column of Table 1. The asymptotic formulas (4) and (6) used for the KN-EO condition, resulted in estimates of K_p and K_n close to those obtained by the optimization method ($K_p = 9.51$ Nm/deg and $K_n = 1{,}035.4$ Nm). The respective

Table 1 Parameter values found by optimization for all the simulated conditions

Parameters	Conditions			
	UP-EO	UP-EC	KN-EO	KN-EC
K_p (Nm/deg)	16.8	16.8	11.0	10.3
K_d (Nms/deg)	4.0	4.0	2.4	2.3
K_n (Nm)	728.7	1045.2	1015.7	1561.5

UP and KN refer to standing and kneeling positions, respectively, in both vision conditions (EO and EC). K_p, K_d and K_n are the proportional, derivative and neural noise gains, respectively model-based spectra are not shown because they follow closely the ones obtained experimentally.

Therefore, summarizing the results of the model-fitting procedures, when the subject went from a standing posture to a kneeling posture the following were observed:

1. The proportional gain K_p decreased by about 34%
2. The derivative gain K_d decreased by about 41%
3. The neural (or torque) noise power increased by about 93%, or alternatively, the noise RMS (root mean square) value increased by about 39%.

As a final investigation, the model parameters were also estimated for the closed-eyes condition, yielding values for K_p and K_d close to those found for open eyes (Table 1). On the other hand, K_n increased 43% in UP and 54% in KN when compared to the EO condition.

Discussion

The present work offers an account of the postural control system behavior under an altered body configuration. During KN there is an absence of proprioceptive input from the foot sole and structures related to the ankle. Therefore, the postural control system in KN has to rely on the available somatosensory inputs coming from structures associated with the knee joint (e.g., thigh muscles), plus the visual and vestibular inputs.

The range of CPx variation was quite far from the boundaries of support in both KN and UP, which implies that the CNS was very probably not activated to change postural strategy either in KN or in UP (Duarte and Zatsiorsky 2002).

The decrease in postural sway in KN (area within the low frequency range in the PSD) could be expected due to the altered biomechanics. For example, the location of the center of mass nearer to the ground (compared to UP) decreases the instability of the biomechanical system. Indeed, the modeling results showed that the effects of the altered biomechanics alone would cause a substantial decrease in the slow components of the CPx in KN. However, this reduction was much greater than that found experimentally (see Fig. 7a for EO condition), suggesting that the nervous system has also changed. This was confirmed by the modeling study, which indicated 34% to 41% changes in K_p, K_d, and K_n (Table 1). The values of K_p, K_d decreased and the value of K_n increased. The latter means that there was an increase in the neural noise power in KN, which could arise from several causes, such as an increase in spinal cord synaptic noise due to changes in presynaptic inhibition (Manjarrez et al. 2005), an increase of the central estimation errors (Kiemel et al. 2002) of the proprioceptive feedback arising from the knee mechanoreceptors and sensory reweighting (Mergner et al. 2001; van der Kooij et al. 2001).

The decrease in K_p and K_d may come from different sources as these parameters encompass the dynamics of three components: sensory feedback, central nervous system and leg muscle (mostly soleus for quiet posture). For example, poorer proprioceptive feedback from the knee may result in lower gains in the transformation of angle variations into neuronal firing rate. The central nervous system itself must be responsible for part of the gain changes, as suggested by the following considerations. From a control theory point of view, the short inverted pendulum associated with the KN position needs a smaller value of K_p to achieve stability. The application of the classical Routh's stability criterion to the system of Fig. 2 gives $K_pG_r > mgd$ as a necessary condition for stability. In the UP-EO condition mgd was 648.0 Nm/rad, while in the KN-EO condition mgd was 343.4 Nm/rad, which indicates that the CNS may set a lower value for K_p in the KN-EO condition. The values K_pG_r/mgd for the UP-EO and KN-EO conditions are 1.48 and 1.83, respectively. Hence, in standing there is a smaller margin for K_p decreases from the natural values than during kneeling in order to maintain stability. To ease the mathematical analysis, we set $K_I = 0$ and $\tau_d = 0$ turning the postural control system into a simple second order system. Its damping ratios (ζ) for the UP-EO and KN-EO conditions resulted 0.80, and 0.87, respectively. This indicates that in both cases the postural control system is slightly underdamped, exhibiting a small tendency for oscillation when the system is perturbed. The corresponding damped natural frequency ω_d for the UP and KN control systems resulted 1.30 and 1.76 rad/s. This indicates that during kneeling a perturbation would cause a faster frequency of oscillation than during standing, which matches one's intuition. The undamped frequency of resonance ω_n for the KN-EO condition was 3.65 rad/s, versus 2.18 rad/s for the UP-EO condition, which justifies the higher power of the CPx power spectrum at larger frequencies for the KN-EO condition when the power spectra for UP and KN are normalized at the lowest frequency value (Fig. 6). Actually, the results obtained from the parameter MV (which reflects fast variations in CPx) suggests an increased muscle activity to maintain the equilibrium under KN (Maki et al. 1990; Krafczyk et al. 1999). An additional possible physiological contribution to the higher relative power at the highest spectral frequencies during KN is the fiber composition of the thigh muscles, which include fast fibers, as compared to the soleus muscle, which practically has only slow fibers. Shorter duration twitches associated with fast fibers lead to torques with higher-frequency content.

It is interesting to note that the model parameters K_p and K_d fitted to the closed-eyes condition (EC) experimental data were similar to those with EO (Table 1), i.e., the proportional and derivative gains stayed practically unchanged when the subjects closed their eyes. On the other hand, the neural noise power increased considerably when the eyes were closed, 43% for the UP position and 54% for the KN position. This could be due to a sensory reweighting mechanism that increased the role played by the vestibular system, which has been reported to be a source of large neural noise (Mergner et al. 2001; van der Kooij et al. 2001). Accordingly, lack of vision caused an increase in the power throughout the spectrum regardless of the postural configuration (see left panels of Fig. 5). These results are in agreement with previous reports for standing subjects (Duarte and Zatsiorsky 2002; Mezzarane and Kohn 2007). The two-way ANOVA results (which showed significant interaction between vision and position) and the profile of the power spectra depicted in Fig. 5c, suggest that the effect of vision was more influential in KN regarding the higher frequency range of postural oscillation (see also Fig. 4c). The observed increase in the MV parameter in the absence of vision is in accordance with previous studies during upright stance (Ramdharry et al. 2006; Baccini et al. 2007; Slobounov et al. 2008).

A more refined separation of the neuro-muscular system parameters that may change their values when posture varies from UP to KN (e.g., the relative importance of proprioceptive, visual and vestibular afferents and the differences in the effectors) would have to be approached by more complex models (van der Kooij et al. 2001; Maurer et al. 2006) as well as complementary experimental approaches. Nevertheless, the rather simple model of PID control of an inverted pendulum was found to be helpful as an aid in the interpretation of the results obtained in this first study of postural control of a kneeling subject.

An open issue is whether the neural noise is higher in KN because the subjects in our experiments were untrained in this postural condition: only two trials with 1 min in each visual condition were performed, probably discarding any learning or training effect (Tarantola et al. 1997). Additional experiments could explore this issue by including training sessions and examining the evolution of the values of CPx and model parameters.

The results obtained here for healthy young subjects could be compared with data obtained from elderly subjects or subjects under rehabilitation physiotherapy for increasing postural stability. Additional posturographic investigation in KN position may provide insights on the contribution of muscles around the knee and axial muscles to body stabilization. This could be achieved by approaches similar to those used in the study of sitting subjects (Genthon and Rougier 2006) or by means of induced fatigue in lumbar extensor muscles (Madigan et al. 2006; Vuillerme et al. 2007).

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