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Stiffness of a wobbling mass models analysed by a smooth orthogonal decomposition of the skin movement relative to the underlying bone

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ABSTRACT

The so-called soft tissue artefacts and wobbling masses have both been widely studied in biomechanics, however most of the time separately, from either a kinematics or a dynamics point of view. As such, the estimation of the stiffness of the springs connecting the wobbling masses to the rigid-body model of the lower limb, based on the *in vivo* displacements of the skin relative to the underling bone, has not been performed yet. For this estimation, the displacements of the skin markers in the bone-embedded coordinate systems are viewed as a proxy for the wobbling mass movement.

The present study applied a structural vibration analysis method called smooth orthogonal decomposition to estimate this stiffness from retrospective simultaneous measurements of skin and intra-cortical pin markers during running, walking, cutting and hopping.

For the translations about the three axes of the bone-embedded coordinate systems, the estimated stiffness coefficients (*i.e.* between 2.3 kN/m and 55.5 kN/m) as well as the corresponding forces representing the connection between bone and skin (*i.e.* up to 400 N) and corresponding frequencies (*i.e.* in the band 10–30 Hz) were in agreement with the literature. Consistently with the STA descriptions, the estimated stiffness coefficients were found subject- and task-specific.

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1. Introduction

The movement of the skin, muscles, and fat relative to the underlying bone is a well-known phenomenon. It has been described, from a kinematics point of view, as the soft tissue artefact (STA). Indeed, this relative movement has a deleterious effect on the joint kinematics estimated from skin markers and motion capture systems (Leardini et al., 2005; Peters et al., 2010). At the same time, from a dynamics point of view, the soft tissue motion, modelled as wobbling masses connected to the rigid-body model of the lower limb, is also recognised to have an effect on the joint kinetics (*i.e.* energy dissipation, torque reduction) (Challis and Pain, 2008; Gruber et al., 1998) during motor tasks involving impacts with the ground.

One key parameter of these wobbling mass models is the stiffness of the springs connecting them to the rigid-bodies. Most of the models of the literature include linear or non-linear springs attached to a wobbling mass that can translate (and eventually rotate) with respect to the bone (Alonso et al., 2007; Gittoes et al., 2006; Gruber et al., 1998; Günther et al., 2003; McLean et al., 2003; Pain and Challis, 2004; Wilson et al., 2006). Identification of the parameters of these wobbling mass models, based on the ground reaction forces, as well as sensitivity analyses have been widely performed (Alonso et al., 2007; Gittoes et al., 2009; Pain and Challis, 2004; Wilson et al., 2006). However, to the best of the author's knowledge, the estimation of the stiffness parameters from the displacements of the skin relative to the underling bone measured *in vivo* by intra-cortical pins has not been performed yet. For this estimation, the displacements of the skin markers in the bone-embedded coordinate systems are viewed as a proxy for the wobbling mass movement.

The objective of this study was to estimate the stiffness matrix of a wobbling mass model, defined as a cluster of lumped masses undergoing translations about the three axes of the bone-embedded coordinate system, by applying a structural vibration analysis method, called smooth orthogonal decomposition (Chelidze and Zhou, 2006) to the simultaneous measurements of skin and intra-cortical pin markers (Benoit et al., 2006; Reinschmidt et al., 1997). In this method, the displacement of the skin markers relative to the underlying bone was modelled as the free undamped vibrations of a dynamical system for which the stiffness matrix can be straightforwardly identified.

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Nomenclature

i	index for segment	λ, $\boldsymbol{\lambda}$	smooth orthogonal value, diagonal matrix of eigenvalues
j	index for marker	ω	circular frequency
k	index for sampled instant of time	$\mathbf{K}, \tilde{\mathbf{K}}, \tilde{K}$	stiffness matrix, stiffness coefficient
n	number of sampled instants of time	M, $\mathbf{M}$	mass, mass matrix
m	number of markers on a segment	$\mathbf{E}$	identity matrix
$\mathbf{v}$, $\mathbf{V}$	STA vector, STA field	$\boldsymbol{\Phi}$	basis vector (*i.e.* marker-cluster geometrical transformation)
$\mathbf{S}$	sample covariance matrix	$\mathbf{F}$	force vector
$\mathbf{D}$	differential operator	a	modal amplitude
f	sampling frequency	$\bar{v}$	displacement of the centroid of the marker-cluster
$\mathbf{U}$	velocity covariance matrix		
$\boldsymbol{\Psi}$	smooth orthogonal vector		

2. Material and methods

2.1. Smooth orthogonal decomposition of the skin movement relative to the underlying bone

The STA vector, $\mathbf{v}_i^j(k)$, was defined to represent the displacement that the skin marker j ($j = 1{:}m_i$) associated with the segment i ($i = 1$ for shank and $i = 2$ for thigh) underwent relative to a relevant bone-embedded coordinate system and a reference position at each discrete time k ($k = 1{:}n$) during the analysed motor task (Dumas et al., 2014a). The STA of all markers on the segment i were represented using the STA field, $\mathbf{V}_i(k)$:

$$\mathbf{V}_i(k) = \begin{pmatrix} \vdots \\ \mathbf{v}_i^j(k) \\ \vdots \end{pmatrix} \quad (1)$$

A sample covariance matrix was computed from this STA field known at every sampled instants of time:

$$\mathbf{S}_i = \frac{1}{n}[\cdots \quad \mathbf{V}_i(k) \quad \cdots][\cdots \quad \mathbf{V}_i(k) \quad \cdots]^T \quad (2)$$

This sample covariance matrix has been used in previous studies to compute, by a proper orthogonal decomposition, the main components of the STA during a running task (Dumas et al., 2014a,b). In the smooth orthogonal decomposition, a differential operator was further introduced:

$$\mathbf{D} = f\begin{bmatrix} -1 & 1 & 0 & \cdots & 0 \\ 0 & -1 & 1 & \cdots & 0 \\ \vdots & \ddots & \ddots & \ddots & 0 \\ 0 & \cdots & 0 & -1 & 1 \end{bmatrix} \quad (3)$$

where f was the sampling frequency.

This differential operator allowed the computation of the other covariance matrix standing for the velocities of the skin markers:

$$\mathbf{U}_i = \frac{1}{n-1}\mathbf{D}[\cdots \quad \mathbf{V}_i(k) \quad \cdots][\cdots \quad \mathbf{V}_i(k) \quad \cdots]^T[\mathbf{D}]^T \quad (4)$$

The smooth orthogonal modes were solutions of the eigen-problem:

$$\mathbf{S}_i[\cdots \quad \boldsymbol{\Psi}_i^l \quad \cdots] = \boldsymbol{\lambda}_i\mathbf{U}_i[\cdots \quad \boldsymbol{\Psi}_i^l \quad \cdots] \quad (5)$$

with $\boldsymbol{\Psi}_i^l$ ($l = 1{:}3m_i$) the smooth orthogonal vectors and $(\omega_i^l)^2 = (\lambda_i^l)^{-1}$ the circular frequencies related to the smooth orthogonal values. Assuming that the STA field was the results of free undamped vibrations of a cluster of lumped masses (*i.e.* each markers having a same mass M_i/m_i), the smooth orthogonal modes are good estimates of the linear normal modes. Therefore, the dynamics of the marker-cluster, as observed from the STA field, was characterized by:

$$\mathbf{K}_i[\cdots \quad \boldsymbol{\Psi}_i^l \quad \cdots] = \mathbf{M}_i[\cdots \quad \boldsymbol{\Psi}_i^l \quad \cdots][\boldsymbol{\lambda}_i]^{-1} \quad (6)$$

with $\mathbf{K}_i$ the stiffness matrix, $\boldsymbol{\lambda}_i$ the diagonal matrix composed of the smooth orthogonal eigenvalues and $\mathbf{M}_i = \frac{M_i}{m_i}\mathbf{E}$ the mass matrix (*i.e.* with $\mathbf{E}$ the identity matrix of dimension $3m_i \times 3m_i$).

Therefore, the stiffness matrix was given by:

$$\mathbf{K}_i = \frac{M_i}{m_i}[\cdots \quad \boldsymbol{\Psi}_i^l \quad \cdots][\boldsymbol{\lambda}_i]^{-1}[\cdots \quad \boldsymbol{\Psi}_i^l \quad \cdots]^{-1} \quad (7)$$

According to the recent descriptions of the STA (Andersen et al., 2012; Benoit et al., 2015; Dumas et al., 2015; Grimpampi et al., 2014) and to the wobbling mass models reported in the literature (Alonso et al., 2007; Bélaise et al., 2016; Challis and Pain, 2008; Gittoes et al., 2009; Gruber et al., 1998; Günther et al., 2003; McLean et al., 2003; Wilson et al., 2006), it was useful to retrieve the stiffness matrix corresponding only to the modes defining the rigid marker-cluster geometrical transformations and more specifically to the marker-cluster translations. This stiffness matrix was given by:

$$\tilde{\mathbf{K}}_i = [\cdots \quad \boldsymbol{\Phi}_i^l \quad \cdots]^T\mathbf{K}_i[\cdots \quad \boldsymbol{\Phi}_i^l \quad \cdots] \quad (8)$$

where $\boldsymbol{\Phi}_i^l$ ($l = 1{:}3$) were the unitary basis vectors built *a priori* (Dumas et al., 2014a) to define the 3 translations of the marker-cluster about the axes of the bone-embedded coordinate system. These basis vectors were:

$$\boldsymbol{\Phi}_i^1 = \begin{pmatrix} \vdots \\ 1/\sqrt{m_i} \\ 0 \\ 0 \\ \vdots \end{pmatrix}, \boldsymbol{\Phi}_i^2 = \begin{pmatrix} \vdots \\ 0 \\ 1/\sqrt{m_i} \\ 0 \\ \vdots \end{pmatrix}, \text{ and } \boldsymbol{\Phi}_i^3 = \begin{pmatrix} \vdots \\ 0 \\ 0 \\ 1/\sqrt{m_i} \\ \vdots \end{pmatrix} \quad (9)$$

Finally, the force vector representing the connection between the bone and skin, applied to each marker j at each discrete time k along each axes of the bone-embedded coordinate system, were given by:

$$\mathbf{F}_i(k) = \mathbf{K}_i\mathbf{V}_i(k) = [\cdots \quad \boldsymbol{\Phi}_i^l \quad \cdots]\tilde{\mathbf{K}}_i\underbrace{[\cdots \quad \boldsymbol{\Phi}_i^l \quad \cdots]^T\mathbf{V}_i(k)}_{\begin{pmatrix} \vdots \\ a_i^l(k) \\ \vdots \end{pmatrix}} \quad (10)$$

Note that the last factors of Eq. (10) matched the definition of the amplitude of the STA (Dumas et al., 2014a), that is to say the projection of the STA field on a given mode:

$$a_i^l(k) = (\boldsymbol{\Phi}_i^l)^T\mathbf{V}_i(k) \quad (11)$$

As the basis vectors $\boldsymbol{\Phi}_i^l$ ($l = 1{:}3$) represented the translations of the marker-cluster, each marker j had the same projected displacement and, therefore, the same force. Moreover, $\bar{v}_i^l(k) = \frac{a_i^l(k)}{\sqrt{m_i}}$ directly corresponded to the projected displacement of the centroid of the marker-cluster about the relevant axis of the bone-embedded coordinate system (*i.e.* $l = 1$ for X axis, $l = 2$ for Y axis, and $l = 3$ for Z axis). The basis vectors $\boldsymbol{\Phi}_i^l$ ($l = 1{:}3$) were also orthogonal and, therefore, the stiffness matrix defining the marker-cluster translations was diagonal (*i.e.* $\tilde{\mathbf{K}}_i = diag(\cdots \quad \tilde{K}_i^l \quad \cdots)$).

2.2. Experimental data

The retrospective data used in this study included the right thigh and shank movements from five trials of a running task (*i.e.* stance phase, from ground contact to take-off) performed by three able-bodied male subjects (Reinschmidt et al., 1997), and from one trial of walking and cutting tasks (*i.e.* stance phase, from ground contact to take-off) and hopping task (*i.e.* during 0.67 s after ground contact) performed by one able-bodied male subject (Benoit et al., 2006). Clusters of markers were attached to intra-cortical pins inserted into the lateral tibial and femoral epicondyles. Both intra-cortical pin markers and skin markers (*i.e.* between 4 and 6 by segments) were tracked using either three high-speed cameras at 200 Hz (Reinschmidt et al., 1997) or tracked using four infrared cameras at 120 Hz

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