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An application of the flexible multibody approach used to estimate human skeleton loading

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Summary. Skeletal loading can be estimated using several approaches. The most common approach is based on utilizing mechanical principles and ground reaction forces as predictors for skeletal loading. This method can be considered as a relatively simple approach since it cannot account for muscle forces. Flexible multibody approach allows for estimating skeletal loading and strains within the bones; once bone flexibility, muscle forces, ground reaction forces and the natural motion of a subject have been accounted for. This paper presents a summary that describes how deformable bodies can be introduced to the standard multibody formulation and explains the benefits and drawbacks. As an example of application, models used to assess tibial strains among two subjects are presented. The results of the multibody simulations are compared to *in vivo* studies, showing acceptable correlation and method performance.

Key words: bone, strain, finite element, simulation, loading, MRI, CT

Introduction

Osteoporosis, accidents and subsequent bone fractures cause suffering on an individual level, as well as an economical burden to society [1, 2]. It has been estimated that between 30,000 to 40,000 osteoporosis-related fractures occur annually and that 400,000 Finnish people have osteoporosis [3]. Between the years of 1998 to 2000, there were approximately 6,000 hip fractures (this amount only accounts for individuals who suffered their first hip fracture) annually in Finland. More than 90% of these accidents happened to people older than 50 years of age [4, 5]. In order to withstand prevalent loading without breaking; while remaining relatively light in weight to allow for locomotion, bones have the ability to adapt their structure to functional loading [6, 7]. Bones are loaded in daily activities by muscles and resist the pull of gravity while accelerating and decelerating body segments [8]. It has been demonstrated that physical activity in the general population strongly affects the amount of weight a skeleton can withstand [9], and therefore, the skeleton is loaded mainly by locomotory actions that impart strains on bones. One method to estimate loading, and thus strains, caused by locomotory actions on the bones is to examine the ground reaction forces registered during these actions [10, 11, 12, 13]. This method is, however, only applicable to the lower body. Nevertheless, it is rarely brought into question if estimating skeletal loading from ground reaction forces is reasonable. Joint angles as well as muscle activity have a great influence on the loading of different bones and should be considered. In addition, diverse bone geometry and mineral content also have a great influence on bone strains. An equal amount of force applied to different bones will lead to different bone strains if the mineral amount and/or geometry of the loaded bones differ.

Since strains are one of the most important stimulants to bone adaptation [14, 15], designing of osteogenic interventions could benefit from the knowledge of bone strains at different cross-sections in a wide range of exercises. *In vivo* bone strain measurements are limited to superficial bone sites [16], and therefore, measuring multiple clinically interesting bone sites is not feasible. Modeling based approaches are expected to provide a reasonable alternative for estimating the skeletal loading and strains at different locations during dynamic movement. Consequently, flexible multibody dynamics is used in this study to estimate bone strains during human walking [17, 18, 19]. The purpose of this paper is to highlight some of the practical challenges affecting the feasibility of the approach as well as to present an outline of the method. Finally, a discussion of other possible modeling approaches, their benefits and drawbacks is included in this study.

Flexibility in multibody applications can be accounted for in a number of ways. The linear theory of elastodynamics can be considered as a traditional approach to account for flexibility. This approach relies on the assumption of small deformations in the flexible bodies. Thus, rigid body simulation is decoupled with the deformation computation. The rigid body simulation is performed to obtain the external, as well as internal, forces acting upon each of the bodies. These forces are later imposed on the finite element model of the body for which deformations, stresses and strains can be obtained [20, 21, 22, 23, 24, 25]. Standard multibody and finite element solvers can be used for this method, which is a great benefit. Additionally, a considerable increase in computational speed can be achieved if the force application points are known *a priori* and do not vary over the duration of the simulation. In such cases, the use of linear finite element analysis is reduced to a single computation of the full model. Strains, stresses and deformations can then be computed for each time step as post-processing. Conversely, the flexibility of the bodies does not affect the multibody simulation behavior, which is the main disadvantage, especially in case of considerable deformations.

Lumped mass formulation is another method that can be used to describe mechanical flexibility [26, 27, 28, 29, 30, 31, 32, 33]. In this formulation, a flexible body is replaced as a set of point masses connected via springs. Using a sufficient amount of springs and masses allows for reasonably accurate mass distribution, within inhomogeneous bodies as well. Similar to the linear theory of elastodynamics, the lumped mass approach does not require that any changes be made in the standard rigid multibody solver. In contrast to the rigid body representation, the performance decays with the rise of disretization precision. Therefore, this method can only be practically applied to beam structures.

The third major method for introducing flexibility into the multibody formulation is the floating frame of reference [34]. Formulation relies on coordinate partitioning so that one set of coordinates is used to describe the flexible body's reference frame in the global coordinate system and another set of coordinates is used to describe the deformation of the body in the local frame of reference. Originally, the deformation of the body was described in a similar fashion to the finite element method, resulting in a remarkable increase in regards to computational effort. However, Shabana and Wehage [35] have developed a solution to this problem by replacing the full finite element models of flexible structures with deformation modes description. This allowed for a reduction in the amount of deformation coordinates (in the range of thousands) to a reasonable amount of modal coordinates, making this method an effective compromise between accuracy and computational effort. Additional details on different flexible multibody formulations can be found in the comprehensive survey of Wasfy and Noor [36].

Materials & methods

The floating frame of reference formulation was chosen to describe strains in the bones described in the authors' recent studies [17, 18, 19, 37]. Global position of a particle, $\mathbf{r}$, located on a deformable body in the floating frame of reference can be described as follows:

$$\mathbf{r} = \mathbf{R} + \mathbf{A}(\bar{\mathbf{u}}_0 + \bar{\mathbf{u}}_f) \tag{1}$$

where $\mathbf{R}$ is the position vector of the local frame of reference, $\bar{\mathbf{u}}_0$ is the vector of location of the particle described in the local frame in the undeformed state and $\bar{\mathbf{u}}_f$ is the vector describing the translation of a particle due to deformation. Modal reduction technique can be applied to $\bar{\mathbf{u}}_f$ coordinates,

$$\bar{\mathbf{u}}_f = \mathbf{\Phi}\mathbf{p} \tag{2}$$

where $\mathbf{\Phi}$ is the matrix that contains assumed deformation modes, and $\mathbf{p}$ is the vector of modal coordinates. Assumed deformation modes needed in the description of deformation can be obtained from the Craig-Bampton [5] modal reduction method. The principle of virtual work can be used to express inertial, elastic and externally applied forces in terms of generalized coordinates. The inertial forces, $\mathbf{F_{iner}}$, of a flexible body can be written as:

$$\mathbf{F_{iner}} = \int_V \rho \ddot{\mathbf{r}}^T dV \tag{3}$$

where ρ is the density of a particle within the body, V is the volume of the flexible body and $\ddot{\mathbf{r}}$ is the acceleration vector of the particle. By applying the concept of virtual displacement, the virtual work done by inertial forces can be expressed as:

$$\delta W_{int} = \int_V \rho \ddot{\mathbf{r}}^T dV \delta\mathbf{r} \tag{4}$$

The virtual displacement of vector $\mathbf{r}$ can be expressed in the form:

$$\delta\mathbf{r} = \frac{\delta\mathbf{r}}{\delta\mathbf{q}}\delta\mathbf{q} = \begin{bmatrix} \mathbf{I} & -\mathbf{A}\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}} & \mathbf{A}\mathbf{\Phi} \end{bmatrix} \delta\mathbf{q} \tag{5}$$

where $\mathbf{A}$ is the rotation matrix, $\mathbf{q}$ is the generalized coordinates vector and $\bar{\mathbf{G}}$ is the transformation matrix coupling first time derivatives of orientation parameters and angular velocities expressed in the local frame of reference. Combining equations (4) and (5) yields:

$$\delta W_{int} = \int_V \rho \ddot{\mathbf{r}}^T \frac{\delta\mathbf{r}}{\delta\mathbf{q}} dV \delta\mathbf{q} = \mathbf{Q}_i^T \delta\mathbf{q} \tag{6}$$

The generalized inertial forces $\mathbf{Q}_i$ can be written in the form:

$$\mathbf{Q}_i^T = \int_V \rho \ddot{\mathbf{r}}^T \frac{\delta\mathbf{r}}{\delta\mathbf{q}} dV \tag{7}$$

Differentiating equation (1) twice leads to a description of the acceleration vector $\ddot{\mathbf{r}}$:

$$\ddot{\mathbf{r}} = \begin{bmatrix} \mathbf{I} & -\mathbf{A}\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}} & \mathbf{A}\mathbf{\Phi} \end{bmatrix} \begin{bmatrix} \ddot{\mathbf{R}} \\ \ddot{\boldsymbol{\theta}} \\ \ddot{\mathbf{p}} \end{bmatrix} + \begin{bmatrix} \mathbf{0} & -\mathbf{A}\tilde{\bar{\boldsymbol{\omega}}}\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}} - \mathbf{A}\dot{\tilde{\bar{\mathbf{u}}}}\dot{\bar{\mathbf{G}}} & \mathbf{A}\tilde{\bar{\boldsymbol{\omega}}}\mathbf{\Phi} \end{bmatrix} \begin{bmatrix} \dot{\mathbf{R}} \\ \dot{\boldsymbol{\theta}} \\ \dot{\mathbf{p}} \end{bmatrix} \tag{8}$$

where $\boldsymbol{\theta}$ is the vector of orientation parameters. Combining equations (5), (7) and (8) yields:

$$\mathbf{Q}_i = \mathbf{M}\ddot{\mathbf{q}} + \mathbf{Q}_v \tag{9}$$

where mass matrix $\mathbf{M}$ is defined as:

$$\mathbf{M} = \int_V \rho \begin{bmatrix} \mathbf{I} & -\mathbf{A}\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}} & \mathbf{A}\boldsymbol{\Phi} \\ \mathbf{G}^T\tilde{\bar{\mathbf{u}}}\mathbf{A}^T & -\mathbf{G}^T\tilde{\bar{\mathbf{u}}}\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}} & \mathbf{G}^T\tilde{\bar{\mathbf{u}}}\boldsymbol{\Phi} \\ \boldsymbol{\Phi}^T\mathbf{A}^T & -\boldsymbol{\Phi}^T\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}} & \boldsymbol{\Phi}^T\boldsymbol{\Phi} \end{bmatrix} dV \tag{10}$$

and the quadratic velocity vector $\mathbf{Q}_v$:

$$\mathbf{Q}_v = \int_V \rho \begin{bmatrix} \mathbf{A}\tilde{\bar{\boldsymbol{\omega}}}\tilde{\bar{\boldsymbol{\omega}}}\bar{\mathbf{u}} + 2\mathbf{A}\tilde{\bar{\boldsymbol{\omega}}}\boldsymbol{\Phi}\dot{\mathbf{p}} - \mathbf{A}\tilde{\bar{\mathbf{u}}}\dot{\bar{\mathbf{G}}}\dot{\boldsymbol{\theta}} \\ \bar{\mathbf{G}}^T\tilde{\bar{\mathbf{u}}}^T\tilde{\bar{\boldsymbol{\omega}}}\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}}\dot{\boldsymbol{\theta}} + \bar{\mathbf{G}}^T\tilde{\bar{\mathbf{u}}}^T\tilde{\bar{\mathbf{u}}}\dot{\bar{\mathbf{G}}}\dot{\boldsymbol{\theta}} - 2\bar{\mathbf{G}}^T\tilde{\bar{\mathbf{u}}}^T\tilde{\bar{\boldsymbol{\omega}}}\boldsymbol{\Phi}\dot{\mathbf{p}} \\ -\boldsymbol{\Phi}\tilde{\bar{\boldsymbol{\omega}}}\tilde{\bar{\mathbf{u}}}\bar{\mathbf{G}}\dot{\boldsymbol{\theta}} - \boldsymbol{\Phi}\tilde{\bar{\mathbf{u}}}\dot{\bar{\mathbf{G}}}\dot{\boldsymbol{\theta}} + 2\boldsymbol{\Phi}\tilde{\bar{\boldsymbol{\omega}}}\boldsymbol{\Phi}\dot{\mathbf{p}} \end{bmatrix} dV \tag{11}$$

Algebraic equations are used for the description of constraints between bodies. Constraint equations are expressed as:

$$\mathbf{C}(q) = \mathbf{0} \tag{12}$$

The concept of virtual work can also be applied to the elastic forces and externally applied forces in a similar manner as with the inertial forces. After the introduction of constraint equations (12), the equation of motion takes the form of a differential algebraic equation and can be formulated as:

$$\mathbf{M}_i\ddot{\mathbf{q}}_i + \mathbf{K}_i\mathbf{q}_i + \mathbf{C}_{qi}^T\boldsymbol{\lambda} = \mathbf{Q}_{ei} + \mathbf{Q}_{vi} \tag{13}$$

where: $\mathbf{Q}_{ei}$ is the vector of generalized forces, $\mathbf{C}_{qi}^T$ is the constraint Jacobian matrix, $\boldsymbol{\lambda}$ is the vector of Lagrange multipliers and generalized reaction forces are represented by the product $\mathbf{C}_{qi}^T\boldsymbol{\lambda}$. Index i points to a single body. Equations (13) and (12) form a set of Differential Algebraic Equations (DAE) which can be converted to ordinary Differential Equations (ODE) to solve for the dynamic response of the multibody system in time domain.

Participants, measurements and the multibody models

Three-dimensional musculoskeletal models with contact description were developed using the LifeMOD (Biomechanics Research Group, Inc., California, USA) plug-in [38] for MSC ADAMS (MSC Software Corporation, California, USA) general multibody software. The skeletons were generated based on the test subjects' weight, height, gender and age. Two Caucasian male subjects volunteered for the experiments. The first subject is 25-years-old, 184cm tall and weighs 89kg. The second subject is a 65-year-old gentlemen, 170cm tall and weighs 65kg. All experiments involving human subjects were conducted in agreement with the Helsinki declaration and were approved by the local ethical committee. All subjects gave their written informed consent prior to the study. Subjects were equipped with an EMG wireless recording set and a suit containing reflective markers. They were then asked to walk with their preferred speed on a level surface. Along the path of the level surface, two 10m long force platforms were installed for independent measurement of the ground reaction forces for both legs. The subjects' motion during one full walking cycle was captured using a stereophotogrammetic method. Muscle activities were then simultaneously measured for several lower limb muscles with surface electromyography.

Table 1. Parameters of the kinematical joints [37].

Joint	Type	Flexion/extension		Inversion/eversion Abduction/adduction		Rotation	
		Stiffness [Nmm/°]	Damping [Nmms/°]	Stiffness [Nmm/°]	Damping [Nmms/°]	Stiffness [Nmm/°]	Damping [Nmms/°]
Ankle	Universal	210	21	10,000	1,000	-	-
Knee	Revolute	270	27	-	-	-	-
Hip	Spherical	700	70	1,500	150	800	80
Lumbar	Revolute	1,000	100	-	-	-	-
Thoracic	Revolute	1,000	100	-	-	-	-
Lower neck	Revolute	1,000	100	-	-	-	-
Upper neck	Fixed	-	-	-	-	-	-
Scapular	Fixed	-	-	-	-	-	-
Shoulder	Universal	700	70	700	70	-	-
Elbow	Revolute	60	6	-	-	-	-
Wrist	Revolute	30	3	-	-	-	-

Among each subject, two types of three-dimensional multibody models were created. The first one is a generic full-body model with a simple closed-loop PID-controlled muscle model [39]. In this model bones were assumed to be rigid. The geometry, mass and inertial properties of the bones were imported from the LifeMOD database, which contains generic shell skeleton models representing the population average. The bones were connected using frictionless kinematic joints with passive stiffness and damping. The omission of friction modeling in the joints is justified because in healthy subjects, this friction is nearly zero. Detailed joint parameters are given in Table 1 [40]. The second type of musculoskeletal models was derived from the generic rigid models. They differed in the sense that they included flexible tibia in first case and in the second case both tibias and both femurs were replaced with their flexible representations

Finite element models

A homogeneous and anisotropic material model, based on values reported in the literature [41], was used for all bones in this study. Young's modulus and the shear elastic modulus of the cortex bone in the longitudinal direction were assumed to be 17 and 10 GPa, respectively. Young's modulus and the shear elastic modulus of the cortex bone were assumed to be transversely isotropic, with values of 5 and 3.5 GPa. Either generic bones obtained from the LifeMOD [17, 19] or a 3D reconstruction from an MRI image stack were used to generate the finite element models [18]. A tetrahedral solid element mesh was used for the bone modeling. An automatic meshing tool from ANSYS (version 11, ANSYS, Inc., Canonsburg, USA) with a fixed element size of 9 mm was applied during the bone model discretization. Bone models were connected with joint centers using massless rigid beams to enable load distribution over the articular cartilage surfaces. The finite element models were subjected to modal analysis to compute natural frequencies and associated natural deformation modes. For the modal analysis, nodes corresponding to joint centers were defined as boundaries for the Craig-Bampton modes. Craig-Bampton modes with an orthonormalization procedure were used to reduce the complexity of the finite element model used in the multibody simulation [42]. For tibia bones, the amount of deformation modes used was between 9 and 14, with corresponding eigenfrequencies ranging from 470 to 20,000 Hz. Generally, from the used deformation modes, 9 influenced the strain results more than 2%. For femur bones, 26 eigenmodes were used in the dynamics simulation. Natural frequencies corresponding to the deformation modes of femurs ranged from 481 to 27,286 Hz. Seven of the modes had an influence above 2%.

Simulation

For each subject, simulation was conducted in two steps. The first step involved inverse dynamics, which aimed at obtaining muscle contraction patterns. During this step, the rigid model was driven by the motion data obtained from experimentation. Consequently, muscles were represented with passive elements, producing no force. As a result of the inverse dynamics, muscle contraction patterns were obtained together with reference body movement. The body movement data was used to determine initial conditions for the forward dynamics simulation.

The second step, forward dynamics simulation, was performed using the model with flexible bones, active muscles and contact models. Contraction patterns of the muscles were used as input signals for the PID controllers of active muscles. Vertical stabilization of the model was achieved with the use of a tracking agent implemented in LifeMOD. The tracking agent uses motion data obtained from inverse dynamics as a reference and applies external torques at the center of mass in the model if necessary. This allows for the compensation of the wobbling masses in the skin, inaccuracy of the mass distribution and errors in the motion capture. The simulation output consists of bone strains at desired locations and ground reaction forces.

A time step of 0.02 seconds was used in all simulations. Foot-to-ground contact was described using an ellipsoid-plane contact model provided by LifeMOD. Contact parameters were determined experimentally and are: damping 20 Ns/mm, stiffness 200 N/mm, static friction coefficient 1, dynamic friction coefficient 0.8, stiction velocity 1 mm/s, and friction velocity 10 mm/s.

Verification of the models was accomplished by comparison of body motion trajectories obtained in the forward dynamics to the inverse dynamics results. Motion pattern accuracy in both cases reached 99.9%. The discrepancies between inverse and forward dynamics simulations are the result of introduced bone flexibility and contact model. It has to be noted, that even though muscle contraction has to be preserved due to the control algorithm, the model is still free to move as a whole, eg. fall down. Thus, body segment trajectories should be verified. The second method of verification consisted of a comparison of the simulated ground reaction forces to the measured values.

Results and discussion

In the presented study's models, kinematics was correctly reproduced with the forward dynamics simulation. Natural human movement was used as a reference for verification of the multibody simulations. Ground reaction forces can be considered the most problematic part of modeling, as they are affected by a large number of parameters, including mass distribution which is difficult to account for. Despite the complexity of modeling of ground reaction forces, they were reproduced to a reasonable extent [17, 18, 19].

In contrast, muscle activity patterns left room for improvement on many occasions. The main problem in modeling muscular force production stems from the redundancy of the musculoskeletal system. This redundancy means that specific joint motion can be

Table 2. The mid-shaft in-plane strains. Literature values from *in vivo* measurements and the values estimated by the models. The principal strains and strain rates are obtained from the anteromedial aspect of the tibial midshaft.

	Strain Magnitude [microstrain]			Strain rate [microstrain/s]		
	Max principal	Min principal	Max shear	Max	Min	Max shear
Lanyon et al. [49]	395	-434	829	-	-4000	-
Burr et al. [8]	437	-544	871	11006	-7183	16162
Milgrom et al. [50]	840	-454	1183	3955	-3306	10303
Milgrom et al. [51]	394	-672	-	4683	-3820	-
Al Nazer et al. [17]	490	-588	1078	3800	-4100	9500
Al Nazer et al. [18]	305	-645	948	4000	-7000	10000

achieved by an infinite combination of muscle activation patterns, differing only in the magnitude of the total force produced by muscles. Energy minimization algorithms can be used to circumvent this problem, as well as a simple muscle force division algorithm based on muscle function and cross-sectional area. Those two approaches however cannot fully represent the antagonistic effect of the muscles under heavy loading. Another part of the problem is that force production of a given muscle cannot be measured (the closest one can get is measuring force output of a muscle group from the tendon, e.g. Achilles tendon and patella tendon *in vivo* force measurements [43, 44]). Recording muscle activity with the use of surface electromyogram (EMG) was utilized as a surrogate for muscular force production in the present studies. It is worth noting sEMG is subject to cross-talk [45] and that force-EMG relationship may not be expected to be linear in any case [46]. Consequently, a reasonable approach for correlating EMG and the modeled force prediction may be to consider just the timing of muscular activity [47]. However, even the timing of muscular activity was incorrect on several occasions, showcasing the difficulty of overcoming the redundancy problem in predicting muscular force outputs.

Nevertheless, for tibial strain estimates, the modeled force output of main muscle groups quadriceps femoris and triceps surae agreed relatively well with the measured EMGs. In terms of the strain estimates, the models gave sound results for tibia (Table 2), showing variation between the two subjects and falling within the measured values *in vivo*. For femur, no comparable data is available in literature, but the timing of the maximal strains agreed with the maximal transverse moment timing measured for instrumented hip implants [48].

Conclusions

Currently, the preferred methods for estimating skeletal loading are based on relationships to ground reaction forces [52, 14, 13]. Unfortunately this approach gives just a rough estimate of bone loading without accounting for bone geometry or the subject's proportions. Alternatively, multibody simulations provide a method to create subject-specific musculoskeletal models. Bone models can be either generic, providing the overall bone proportions and sizes, or they can be reconstructed from medical imaging data to fully match the subject. Material models of the bones can be based on averaged literature values or obtained from mathematical dependencies of elastic properties on computed tomography data [53, 54]. Furthermore, use of different material models for cortical and trabecular bones can enhance model accuracy. Additional enhancement can be achieved with the introduction of an inhomogeneous material model.

Initial strain modeling results have given new insight into bone loading research. Clinically interesting, yet inaccessible, *in vivo* sites can now be investigated. Strain results at accessible bone surfaces obtained from flexible multibody simulations comply with the measurements obtained *in vivo* or *in situ*. This encourages the usage of the method to the previously unavailable bone sites. However, plenty of research is still needed to provide reliable validation data for the multibody simulations, as well as to provide accurate material models.

Bone strains play an important role in bone remodeling. Thus, studying bone strains at various locations within different bones can provide more knowledge of how specific exercises and physical activities influence different bone formation and resorption. It seems plausible that one may theorize an exercise that causes high ground reaction forces, while causing only minimal loading on one or another bone. On the other hand, verifying the influence of exercise on bones by examining bone strains would not lead to such inaccurate conclusions. In conclusion, the presented flexible multibody dynamics appears feasible for modeling skeletal loading. Currently, the Lappeenranta-Jyväskylä research team investigates typical gym exercises in terms of bone strains utilizing the flexible multibody approach. The results of the studies are expected to provide more insight into how different exercises affect skeletal system components, and thus allow providing guidelines for exercise equipment design.

References

[1] J. A. Stevens and S. Olson, "Reducing falls and resulting hip fractures among older women," *MMWR Recomm Rep.*, vol. 49, pp. 3–12, Mar 31 2000.

[2] G. Ortiz-Luna, P. Garcia-Hernandez, and J. A. Tamayo-Orozco, "Treatment options for osteoporosis and decision making criteria: 2009," *Salud publica de Mexico*, vol. 51 Suppl 1, pp. S114–25, 2009.

[3] S. L. Duodecim, "Osteoporoosi," tech. rep., Suomalainen Lääkäriseura Duodecim, 2008.

[4] R. Sund, "Lonkkamurtumien ilmaantuvuus suomessa 1998-2002," *Duodecim; lääketieteellinen aikakauskirja*, vol. 122, no. 9, pp. 1085–1091, 2006.

[5] P. Kannus, S. Niemi, J. Parkkari, M. Palvanen, I. Vuori, and M. Järvinen, "Nationwide decline in incidence of hip fracture," *Journal of bone and mineral research : the official journal of the American Society for Bone and Mineral Research*, vol. 21, pp. 1836–1838, Dec 2006.

[6] H. M. Frost, "Muscle, bone, and the utah paradigm: a 1999 overview," *Medicine and science in sports and exercise*, vol. 32, pp. 911–917, May 2000.

[7] H. M. Frost, "Bone's mechanostat: a 2003 update," *The anatomical record.Part A, Discoveries in molecular, cellular, and evolutionary biology*, vol. 275, pp. 1081–1101, Dec 2003.

[8] D. B. Burr, C. Milgrom, D. Fyhrie, M. Forwood, M. Nyska, A. Finestone, S. Hoshaw, E. Saiag, and A. Simkin, "In vivo measurement of human tibial strains during vigorous activity," *Bone*, vol. 18, pp. 405–410, May 1996.

[9] T. M. Mikkola, S. Sipilä, T. Rantanen, H. Sievänen, H. Suominen, J. Kaprio, M. Koskenvuo, M. Kauppinen, and A. Heinonen, "Genetic and environmental influence on structural strength of weight-bearing and non-weight-bearing bone: a twin study," *Journal of Bone and Mineral Research*, vol. 23, pp. 492–498, Apr 2008.

[10] T. Jämsä, A. Vainionpää, R. Korpelainen, E. Vihriälä, and J. Leppäluoto, "Effect of daily physical activity on proximal femur," *Clin.Biomech.(Bristol, Avon)*, vol. 21, pp. 1–7, Jan 2006.

[11] A. Vainionpää, R. Korpelainen, E. Vihriälä, A. Rinta-Paavola, J. Leppäluoto, and T. Jämsä, "Intensity of exercise is associated with bone density change in premenopausal women," *Osteoporosis International*, vol. 17, no. 3, pp. 455–463, 2006.

[12] A. Vainionpää, "Bone adaptation to impact loading : significance of loading intensity," 2007. Note: Tiivistelmä: Luun mukautuminen liikuntaan : kuormituksenintensiteetin merkitys; Note: also available online (ISBN 978-951-42-8493-9 (PDF))

[13] C. H. Turner and A. G. Robling, "Designing exercise regimens to increase bone strength," *Exercise and sport sciences reviews*, vol. 31, pp. 45–50, Jan 2003.

[14] C. H. Turner, "Three rules for bone adaptation to mechanical stimuli," *Bone*, vol. 23, pp. 399–407, Nov 1998.

[15] L. E. Lanyon, "Functional strain in bone tissue as an objective, and controlling stimulus for adaptive bone remodelling," *Journal of Biomechanics*, vol. 20, no. 11-12, pp. 1083–1093, 1987.

[16] S. J. Hoshaw, D. P. Fyhrie, Y. Takano, D. B. Burr, and C. Milgrom, "A method suitable for in vivo measurement of bone strain in humans," *Journal of Biomechanics*, vol. 30, pp. 521–524, May 1997.

[17] R. A. Nazer, A. Kłodowski, T. Rantalainen, A. Heinonen, H. Sievänen, and A. Mikkola, "Analysis of dynamic strains in tibia during human locomotion based on flexible multibody approach integrated with magnetic resonance imaging technique," *Multibody System Dynamics*, vol. 20, pp. 287–306, 2008.

[18] R. A. Nazer, T. Rantalainen, A. Heinonen, H. Sievänen, and A. Mikkola, "Flexible multibody simulation approach in the analysis of tibial strain during walking," *Journal of Biomechanics*, vol. 41, no. 5, pp. 1036–1043, 2008.

[19] A. Kłodowski, J. Kulmala, A. Mikkola, H. Sievänen, and A. Heinonen, "A dynamic computer simulation of tibial strains and joint forces during knee flexion and extension exercise," in *The First Joint International Conference on Multibody System Dynamics*, (Lappeenranta, Finland), Lappeenranta University of Technology, May 25 - 27, 2010 2010.

[20] A. G. Erdman and G. N. Sandor, "Kineto-elastodynamics – a review of the state of the art and trends," *Mechanism and Machine Theory*, vol. 7, no. 1, pp. 19 – 33, 1972.

[21] G. Lowen and C. Chassapis, “The elastic behavior of linkages: An update,” *Mechanism and Machine Theory*, vol. 21, no. 1, pp. 33 – 42, 1986.

[22] G. G. Lowen and W. G. Jandrasits, “Survey of investigations into the dynamic behavior of mechanisms containing links with distributed mass and elasticity,” *Mechanism and Machine Theory*, vol. 7, no. 1, pp. 3 – 17, 1972.

[23] D. A. Turcic and A. Midha, “Dynamic analysis of elastic mechanism systems. part i: Applications,” *Journal of Dynamic Systems, Measurement, and Control*, vol. 106, no. 4, pp. 249–254, 1984.

[24] D. A. Turcic, A. Midha, and J. R. Bosnik, “Dynamic analysis of elastic mechanism systems. part ii: Experimental results,” *Journal of Dynamic Systems, Measurement, and Control*, vol. 106, no. 4, pp. 255–260, 1984.

[25] R. C. Winfrey, “Elastic link mechanism dynamics,” *Journal of Engineering for Industry*, vol. 93, no. 1, pp. 268–272, 1971.

[26] P. C. Hughes, “Dynamics of a chain of flexible bodies,” *Journal of the Astronautical Sciences*, vol. 27, pp. 359–380, 1979.

[27] T. J. R. Hughes and J. Winget, “Finite rotation effects in numerical integration of rate constitutive equations arising in large-deformation analysis,” *Int. J. Numer. Meth. Engng.*, vol. 15, no. 12, pp. 1862–1867, 1980.

[28] R. L. Huston, “Multi-body dynamics including the effects of flexibility and compliance,” *Computers & Structures*, vol. 14, no. 5-6, pp. 443 – 451, 1981.

[29] R. L. Huston, “Computer methods in flexible multibody dynamics,” *Int. J. Numer. Meth. Engng.*, vol. 32, no. 8, pp. 1657–1668, 1991.

[30] R. L. Huston and Y. Wang, “Flexibility effects in multibody systems, in computer-aided analysis of rigid and flexible mechanical systems,” *Proceedings of the NATO Advanced Study Institute*, pp. 351–376, 1994.

[31] J. W. Kamman and R. L. Huston, “Multibody dynamics modeling of variable length cable systems,” *Multibody System Dynamics*, vol. 5, pp. 211–221, 2001.

[32] W. Raman-Nair and R. Baddour, “Three-dimensional dynamics of a flexible marine riser undergoing large elastic deformations,” *Multibody System Dynamics*, vol. 10, pp. 393–423, 2003.

[33] W. O. Schiehlen and J. Rauh, “Modeling of flexible multibeam systems by rigid-elastic superelements,” *Revista Brasiliera de Ciencias Mecanicas*, vol. 2, pp. 151–163, 1986.

[34] A. A. Shabana, *Dynamics of multibody systems.* Cambridge :: Cambridge University Press, 2005.

[35] A. A. Shabana and R. A. Wehage, “Coordinate reduction technique for transient analysis of spatial substructures with large angular rotations,” *Journal of Structural Mechanic*, 1983.

[36] T. M. Wasfy and A. K. Noor, “Computational strategies for flexible multibody dynamics,” *Applied Mechanics Reviews*, vol. 56, no. 6, pp. 553–613, 2003.

[37] A. Kłodowski, T. Rantalainen, A. Mikkola, H. Ari, and H. Sievänen, “Flexible multibody approach in forward dynamic simulation of locomotive strains in human skeleton with flexible lower body bones,” *Multibody System Dynamics*, pp. 1–15, 2011.

[38] I. B. R. Group, “Manual lifemod biomechanics modeler,” tech. rep., 2005.

[39] C. U. Biomechanics Research Group, Inc., *LifeMOD online user’s manual*, 2007.

[40] K. Amankwah, R. J. Triolo, and R. Kirsch, “Effects of spinal cord injury on lower-limb passive joint moments revealed through a nonlinear viscoelastic model,” *Journal of rehabilitation research and development*, vol. 41, pp. 15–32, Jan-Feb 2004.

[41] X. N. Dong and X. E. Guo, “The dependence of transversely isotropic elasticity of human femoral cortical bone on porosity,” *Journal of Biomechanics*, vol. 37, pp. 1281–1287, Aug 2004.

[42] R. R. Craig and M. C. C. Bampton, “Coupling of substructures for dynamic analysis,” *American Institute of Aeronautics and Astronautics Journal*, vol. 6, no. 7, pp. 1313–1319, 1968.

[43] P. V. Komi, S. Fukashiro, and M. Järvinen, “Biomechanical loading of achilles tendon during normal locomotion,” *Clinics in sports medicine*, vol. 11, pp. 521–531, Jul 1992.

[44] T. Finni, P. V. Komi, and V. Lepola, “In vivo human triceps surae and quadriceps femoris muscle function in a squat jump and counter movement jump,” *European journal of applied physiology*, vol. 83, pp. 416–426, Nov 2000.

[45] C. A. Byrne, G. M. Lyons, A. E. Donnelly, D. T. O’Keeffe, H. Hermens, and A. Nene, “Rectus femoris surface myoelectric signal cross-talk during static contractions,” *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology*, vol. 15, pp. 564–575, Dec 2005.

[46] C. Disselhorst-Klug, T. Schmitz-Rode, and G. Rau, “Surface electromyography and muscle force: limits in semg-force relationship and new approaches for applications,” *Clinical biomechanics (Bristol, Avon)*, vol. 24, pp. 225–235, Mar 2009.

[47] D. D. Anderson, B. M. Hillberry, D. Teegarden, W. R. Proulx, C. M. Weaver, and W. Yoshikawa, “Biomechanical analysis of an exercise program for forces and stresses in the hip joint and femoral neck,” *Journal of Applied Biomechanics*, vol. 12, pp. 292–312, 1996.

[48] G. Bergmann, H. Kniggendorf, F. Graichen, and A. Rohlmann, “Influence of shoes and heel strike on the loading of the hip joint,” *Journal of Biomechanics*, vol. 28, pp. 817–827, Jul 1995.

[49] L. E. Lanyon, W. G. Hampson, A. E. Goodship, and J. S. Shah, “Bone deformation recorded in vivo from strain gauges attached to the human tibial shaft,” *Acta Orthopaedica Scandinavica*, vol. 46, pp. 256–268, May 1975.

[50] C. Milgrom, A. Finestone, Y. Levi, A. Simkin, I. Ekenman, S. Mendelson, M. Millgram, M. Nyska, N. Benjuya, and D. Burr, "Do high impact exercises produce higher tibial strains than running?," *British journal of sports medicine*, vol. 34, pp. 195–199, Jun 2000.

[51] C. Milgrom, D. R. Radeva-Petrova, A. Finestone, M. Nyska, S. Mendelson, N. Benjuya, A. Simkin, and D. Burr, "The effect of muscle fatigue on in vivo tibial strains," *Journal of Biomechanics*, vol. 40, no. 4, pp. 845–850, 2007.

[52] R. T. Whalen, D. R. Carter, and C. R. Steele, "Influence of physical activity on the regulation of bone density," *Journal of Biomechanics*, vol. 21, no. 10, pp. 825–837, 1988.

[53] Y. Rho, J., C. Hobatho, M., and B. Ashman, R., "Relations of mechanical properties to density and ct numbers in human bone," *Med. Eng. Phys.*, vol. 17, no. 5, pp. 347–355, 1995.

[54] B. Helgason, E. Perilli, E. Schileo, F. Taddei, S. Brynjólfsson, and M. Viceconti, "Mathematica relationships between bone density and mechanical properties: A literature review," *Clinical Biomechanics*, vol. 23, pp. 135–146, 2008.

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