2007-01-2504

Assessing the Importance of Motion Dynamics for Ergonomic Analysis of Manual Materials Handling Tasks using the AnyBody Modeling System

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ABSTRACT

Most current applications of digital human figure models for ergonomic assessments of manual tasks focus on the analysis of a static posture. Tools available for static analysis include joint-specific strength, calculation of joint moments, balance maintenance capability, and low-back compression or shear force estimates. Yet, for many tasks, the inertial loads due to acceleration of body segments or external objects may contribute significantly to internal body forces and tissue stresses. Due to the complexity of incorporating the dynamics of motion into analysis, most commercial software packages used for ergonomic assessment do not have the capacity to include dynamic effects. Thus, commercial human modeling packages rarely provide an opportunity for the user to determine if a static analysis is sufficient. The goal of this paper is to quantify the differences between a static and dynamic analysis of a materials handling task using the AnyBody modeling system to include the effects of motion. The feasibility and tractability of performing dynamical ergonomic analysis with the AnyBody model is assessed for the analysis of an asymmetric lifting task. Comparisons between low back moments, compression and shear forces for dynamic and static analyses are presented.

INTRODUCTION

Digital human figure models (DHM) are used for a variety of applications spanning job safety evaluation (including biomechanical, postural, and strength analysis), reach/space accommodation, vision capability and occlusion, and task visualization. Accurate assessment of operator job safety and performance is important to reduce worker injuries and the associated costs. The U.S. Bureau of Labor and Statistics (BLS) reported in 2004 that 5.1 million workers were classified under 'Manual Moving Occupations' (BLS 2006-2007) and reported 173,400 cases (19.5 reported incidences per 10,000 full-time workers) of acute overexertion solely due to lifting. The National Research Council in the United States presented conservative estimates of costs (defined by compensation, lost wages, and lost productivity) in 1999 associated with musculoskeletal disorders to be between $45 and $54 billion annually. Stewart et al. (2003) reported a value of $31.4 billion dollars in 2002 of total productive time lost due to back or unspecified musculoskeletal pain from a random sampling of 28,902 working adults.

Proactive ergonomics using DHM has the potential to reduce the number of work-related injuries by allowing analysts to identify potentially hazardous tasks early in the design of a product or manufacturing facility. Ergonomic assessment tools in DHM for manual material-handling tasks include joint strength capability (i.e., University of Michigan 3D Static Strength Program), the NIOSH lifting index, low back compression, extreme postural analysis, balance maintenance capability, and fatigue assessments. All of these tools require as input a postured figure representing an operator performing the task being analyzed.

Many of the assessment tools can be used in three analysis modes: static, quasi-static, or dynamic. The static analysis analyzes a single posture assumed to be associated with the greatest injury risk or tissue stress during the task. The quasi-static analysis applies a static analysis at multiple time steps of a motion. Inertial effects are neglected, effectively setting the velocity and acceleration components of the body during each frame to zero. The dynamic mode of analysis includes all the aspects of motion in the calculation of joint forces and internal stresses, including the effects introduced by changing velocity and acceleration components.

Most examples of incorporating the dynamics of motion into ergonomic analysis are specific case studies or simulations, often performed in an academic context rather than in industry (Anderson and Pandy 2001, Hooper et al. 1998, Ren et al. 2006). The advantage of a dynamic analysis over static computations depends on task characteristics. For sagittal plane lifting, Dysart and Woldstad (1996) and McGill and Norman (1985) reported that dynamic analyses were significantly more accurate than the quasi-static equivalent in the prediction of external joint torques. The initiation of protective stepping due to a balance disturbance has also been found to be more accurately predicted by a dynamic analysis (Pai et al. 1998). Other studies have not found important dynamic effects in lifting tasks (Zhang et al. 2003). The dependence of the importance of dynamic effects on task characteristics means that it is not, in principle, straightforward for an ergonomist to determine if a task analysis requires a dynamic analysis or conversely if a static analysis will suffice.

The barriers to incorporating dynamic analysis as a routine part of DHM analyses are considerable. Currently, DHM software is not capable of simulating task motions with sufficiently accurate velocities and accelerations. A dynamic analysis consequently requires motion data from a human performing the task of interest. The complexity and cost of acquiring motion capture data, and processing it sufficiently to use for dynamic analysis, limits the application of dynamic analysis to relatively few tasks. A second issue is the availability of inverse-dynamics software capable of computing the internal joint forces and moments associated with a measured motion. The most widely used DHM software systems, such as Jack and Safework, lack built-in inverse-dynamics capability. However, newer software systems, such as AnyBody (Damsgaard et al. 2006), are making these computations available for ergonomics applications.

A third challenge is the lack of analysis tools and tissue or joint-level criteria sufficient for making judgments about the suitability of tasks using the results provided by a dynamic analysis. For example, the AnyBody system is capable of predicting muscle forces using a wide range of potential optimization criteria to resolve the muscle redundancy problem. However, no general-purpose optimality criterion has yet been developed (see, for example, Damsgaard et al. 2006 and Dickerson 2005), so the choice for any particular analysis is not yet apparent. Moreover, criteria for determining whether a particular task is "safe" based on tissue-level stresses are available for only a small number of tissues and loading regimes (e.g., lower back motion segments in compression).

Nonetheless, the fact that dynamics can be consequential in some industrial tasks of ergonomic interest means that research should be directed toward the eventual implementation of routine dynamic analysis in human modeling systems used for ergonomics. The goal of this paper is to explore the feasibility and requirements of incorporating dynamic analysis into commercially available ergonomic analysis tools. A three-dimensional lifting task is analyzed using the AnyBody human modeling system and motion-capture data from the Human Motion Simulation Laboratory at the University of Michigan. Requirements for data transformation, necessary additional information beyond the traditional quasi-static analysis, and required assumptions are addressed.

METHODS

DATA COLLECTION

The human motion data used for analysis were gathered in the Human Motion Simulation (HUMOSIM) laboratory at the University of Michigan as part of a larger study (Wagner et al. 2006). Participants moved boxes and cylindrical objects with a range of weights between pickup and delivery locations while their whole-body motions were recorded. Data for a single trial (Figure 1) with a male participant lifting a 4.54 kg load from a shelf height of 0.967 m are presented here. The participant had an age of 23 years, stature of 1.824 m, body mass of 84.55 kg, and a BMI of 25.11 kg/m^2.

A six-camera Qualisys Proreflex 240-MCU passive optical motion tracking system was used to capture kinematic data at 50 Hz. Foot switches affixed to the ball and heel of the foot inside the shoe of the participant were used to collect heel and toe ground contact times. Two AMTI force plates at the pickup location recorded ground reaction forces. Pressure switches on each shelf were sampled to determine load pickup and delivery times. All analog signals were sampled at 500 Hz. Body segment masses were computed using methods similar to Pataky et al. (2003).

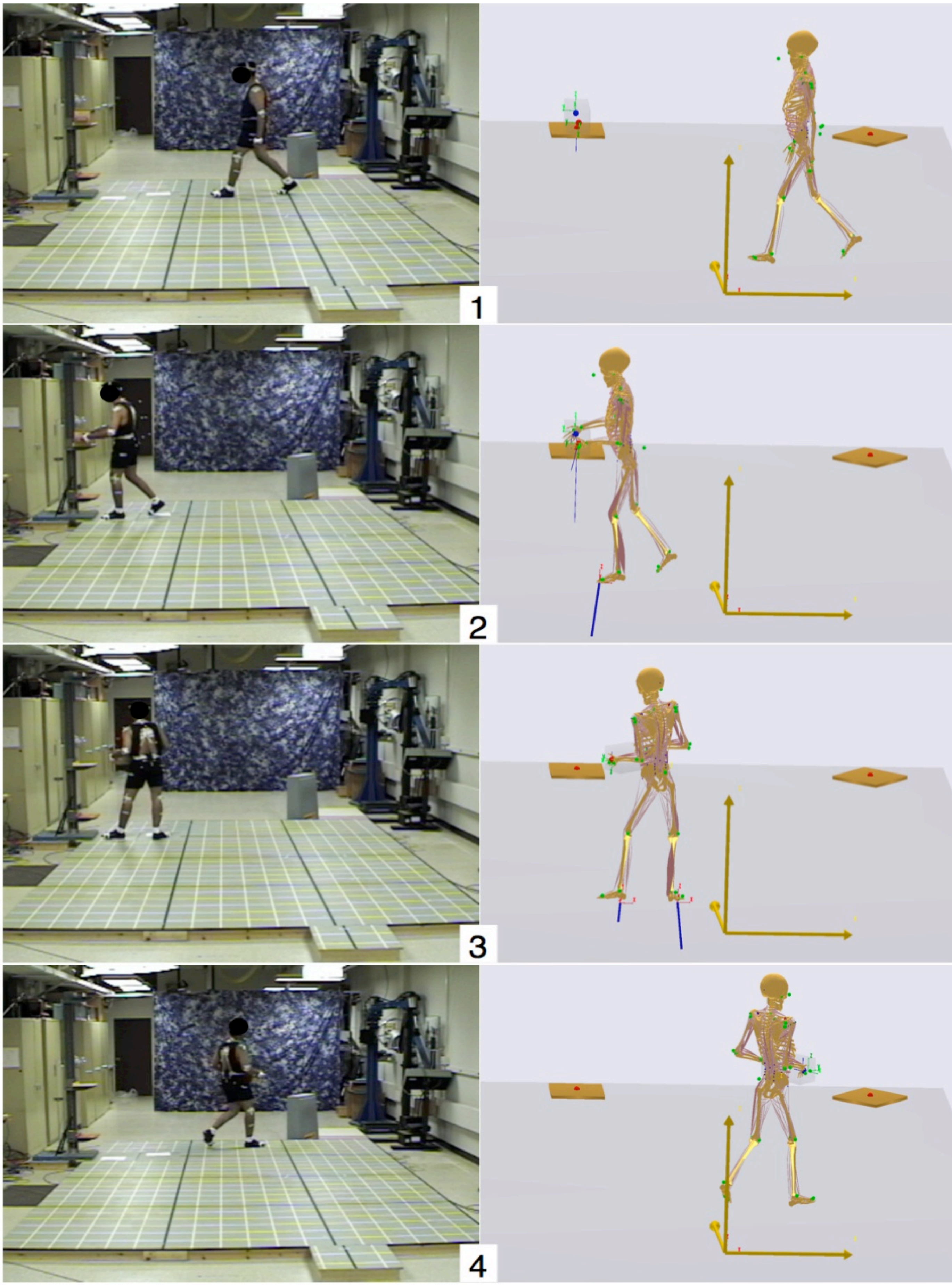

Figure 1. Step progression for the load pickup trial (video and model) used for analysis. 1 - Walking toward the pickup location; 2,3 - Load pickup; 4 - Walking with load toward the delivery location.

ANYBODY MODELING SYSTEM

The AnyBody Modeling System is computer software designed for constructing detailed models of the musculoskeletal system. The mathematical and mechanical methods of the system were described in detail in (Damsgaard et al. 2006). The system is based on inverse dynamics. It presumes that muscles are recruited according to a minimum fatigue criterion and is capable of simulating the force in every muscle and reactions in every joint and external support condition for prescribed movements and external loads. The system is fully dynamic in the sense that body forces from accelerations and gravity are included in the analysis.

In the AnyBody modeling system, the user creates the model of the problem to be investigated. Such a model is termed an application. Unlike traditional digital manikins the human model is not an integrated part of the system. However, a public-domain repository of human body models is available and eliminates the need to build the human anatomy over again for each application. The human model is scaleable, so essentially the same human model in different scaled forms is used in multiple applications.

This current human model comprises approximately 500 individually activated muscles. However, it also comes in a form in which the muscles are replaced by joint moment providers, thus enabling the model to compute joint moments instead of muscle forces.

The current analysis was based on modifications of a standard application from the model repository called FreePostureMove. The manikin model is a 42-degree of freedom (DOF) whole body linkage driven by a combination of angular and positional constraints. Model implementation is presented in three categories: Kinematic definition, anthropometric scaling, and kinetic environment/model reactions.

TRANSITION LIFTING ANYBODY MODEL DEVELOPMENT

Kinematic Model Definition

Joint center and surface landmark positions (referred here as anatomical positions) for this experiment (Figure 2) were calculated from the optical marker data using an approach similar to that described in Wagner et al. (2005). A combination of anatomical positions and joint angles derived from data were used to drive the AnyBody model. Body-fixed reference frames were defined for each body segment from marker positions and body landmarks. A local transition rotation matrix was calculated for adjoining segment reference frames. Joint angles were calculated using the corresponding local transition rotation matrix and the respective rotation sequence defined by the FreePostureMove model. Selected manikin landmarks were constrained to a combination of the {x,y,z} coordinate of an anatomical landmark in the reference frame locally defined by a model body segment. Table 1 lists the model DOF and the respective kinematic drivers.

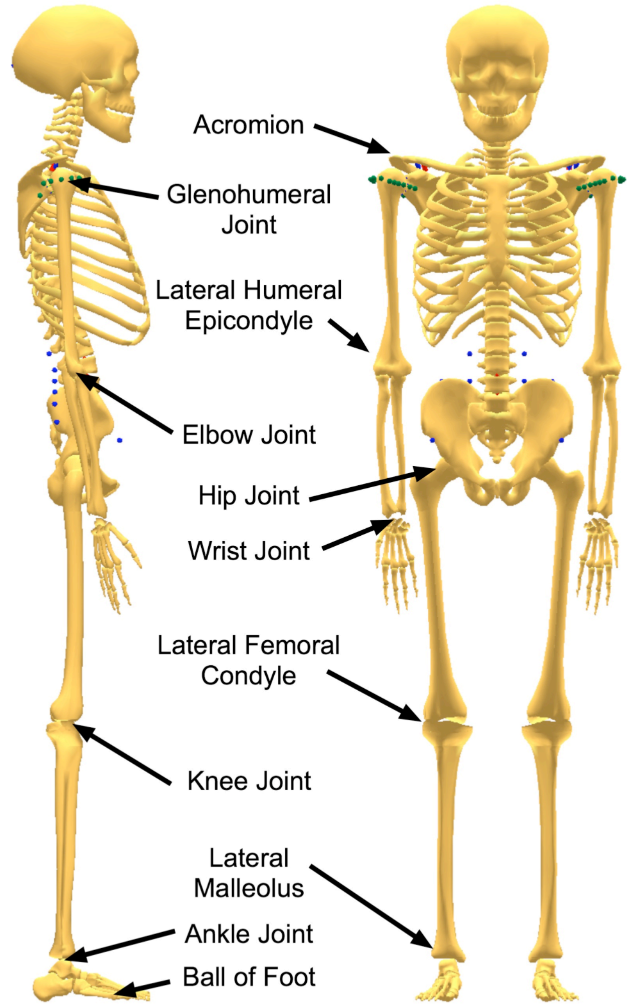

Figure 2. Anatomical positions used to drive the AnyBody manikin model. Symmetric left side anatomical positions are not shown.

Table 1. AnyBody model degrees of freedom and respective kinematic drivers

Manikin Segment	Joint	DOF	Single DOF Definition	Kinematic Driver (Positional or Angular)
Pelvis	Global Reference	6	Position {x,y,z} Rotation {θ,ϕ,ψ}	Positional Angular
Thorax	Thorax-Lumbar Spine	3	Lat. Bending Rotation Extension	Angular
Neck	Cervical Spine	1	Extension	Angular
Clavicle	Sterno-Clavicular	3*	Protraction Elevation Axial Rot.	Angular
Upper Arm	Gleno-humeroid	3	Abduction Flexion External Rot.	Positional
ForeArm	Elbow	2	Flexion Pronation	Positional Angular
Hand	Wrist	2	Flexion Abduction	Angular
Thigh	Hip	3	Flexion Abduction External Rot.	Positional
Shank	Knee	1	Flexion	Positional
Foot	Ankle	2	PlantarFlexion Eversion	Positional

*Angles are fixed at nominal values and not dependent upon the data

Anthropometric Model Scaling

Scaling the model manikin to accurately represent the measured participant is a critical determinant of the accuracy of the results. For example, incorrectly defining limb segment lengths creates errors in the moments due to distal masses or applied forces. Incorrect definition of COM positions can yield inaccurate moment arms due to gravity and segment accelerations. These factors can lead to unrealistic internal body moments (joint torques) and/or forces (muscles exertions) calculated by the inverse dynamics analysis to drive the input motion. Inaccuracy in body segment parameter estimates (masses, mass moments of inertia) can have similarly important effects (Pataky et al. 2003).

The anthropometric scaling of the AnyBody manikin used for analysis was derived from the AnyBody repository function ScalingUniform. Limb lengths and selected body dimensions were geometrically scaled to subject specific input values (Table 2). The necessary 16 segment masses and additional inertial properties to scale the AnyBody manikin were defined by scaling the segment parameters of a mid-size male (Table 3). Segment masses for the thigh, the combined shank and foot, the upper arm, and the combined forearm and hand were measured for the participant using the method described by Pataky (2003) for comparison with the scaled segment values. However, out of plane deviation for the thigh and upper arm during data collection prevented an accurate estimation of those body segment masses. The measured limb masses for the combined shank and foot and combined forearm and hand segments were estimated to be 6.550% and 2.391% of body mass respectively. The segment distribution implemented in ScalingUniform underestimates these masses for the combined shank and foot mass by 0.380 kg and the combined forearm and hand mass by 0.161 kg.

Table 2. AnyBody inputs for model manikin geometric scaling.

Geometric Scaling Measure	ScalingUniform Values (m)
Thigh Length:	0.4091
Shank Length:	0.4528
Foot Length:	0.2106
Pelvis Width:	0.1520
Head Height:	0.16
Trunk Height:	0.6674
Upper Arm Length:	0.3117
Fore Arm Length:	0.2931
*Trunk Width:	0.421

*not available in the ScalingUniform repository implementation; implemented for this model.

Table 3. AnyBody inputs for model manikin mass distribution scaling.

Mass Scaling Measure	ScalingUniform Values (% Body Mass)
Lower Lumbar Spine	4.08
Upper Lumbar Spine	4.62
Lower Thoracic Spine	5.09
Upper Thoracic Spine	6.51
Lower Cervical Spine	0.51
Upper Cervical Spine	0.43
Pelvis	14.2
Clavicle	2.37
Upper Arm	2.8
Lower Arm	1.6
Hand	0.6
Head	8.1
Thigh	10.0
Shank	4.65
Foot	1.45
Ball of Foot	0.0

Kinetic Environment/Model Reactions

Force Plates —Each foot interacted with only one force plate during the trial, so the two force plates were modeled as spatially translating resultant forces applied to the feet of the model. The Center of Pressure (COP) location calculated for each plate defined the point of force application. The resultant Fx, Fy, and Fz forces measured by the force plate defined the direction and magnitude of the applied force. The force plate implementation in AnyBody was derived from the Gait3D application available in the model repository (Damsgaard et al. 2006).

The model is capable of exchanging forces with the environment either as measured and subsequently applied forces or as computed forces resulting from rigid boundary conditions such as the foot contact with the floor. In the latter case, the result depends entirely on the accelerations and mass properties of the individual segments of the body model and how accurately these have been measured. Accelerations in particular are difficult to measure accurately; so measured ground reaction forces are preferable in cases where dynamics plays a significant role. An inverse dynamics model does not work if rigid boundary conditions are completely absent because inevitable discrepancies between the externally applied forces and the masses and accelerations of the model will make it impossible to solve the equilibrium equations. Thus, it is necessary to equip the model with at least one boundary condition for each spatial degree of freedom to absorb differences between the applied forces and the imposed movement of the model. In the present case, this boundary condition is applied at the pelvis. This is advantageous because the spine is rooted at the pelvis as an open chain and likewise the two lower extremities. The open chains are statically determinate and their equilibriums hence depend only on the loads applied at the open ends and not on forces that may be transmitted to the pelvis through the boundary condition.

Lifted Object —The load lifted by the study participant was geometrically modeled as a rectangular prism with equivalent dimensions (0.295 x 0.2 x 0.186 m). The inertial properties of the lifted object were modeled as a point mass located at the geometric center of the box.

TEST CONDITIONS (SIMULATION)

The lifting analyses are presented over a 4 second span centered about the pickup time defined by the pressure switch located at the pickup shelf (Figure 3). A total of six steps were taken during the four seconds presented here; three steps proceeding and three steps succeeding the pickup. The quasi-static simulation was performed in the AnyBody environment by reformulating the dynamic simulation such that the time to complete the transfer task was multiplied by a factor of 1000, effectively negating any inertial effects. The only difference between the dynamic and quasi-static simulations was due to the extension of the time over which the simulations occurred, which reduced the accelerations to negligible levels. Inverse dynamics analyses were calculated for both the dynamic and quasi-static cases. The moments at the low back, specifically those simulated at the L5-Sacrum joint, and the low back shear and compression forces observed at the same location are presented for the dynamic and quasi-static cases.

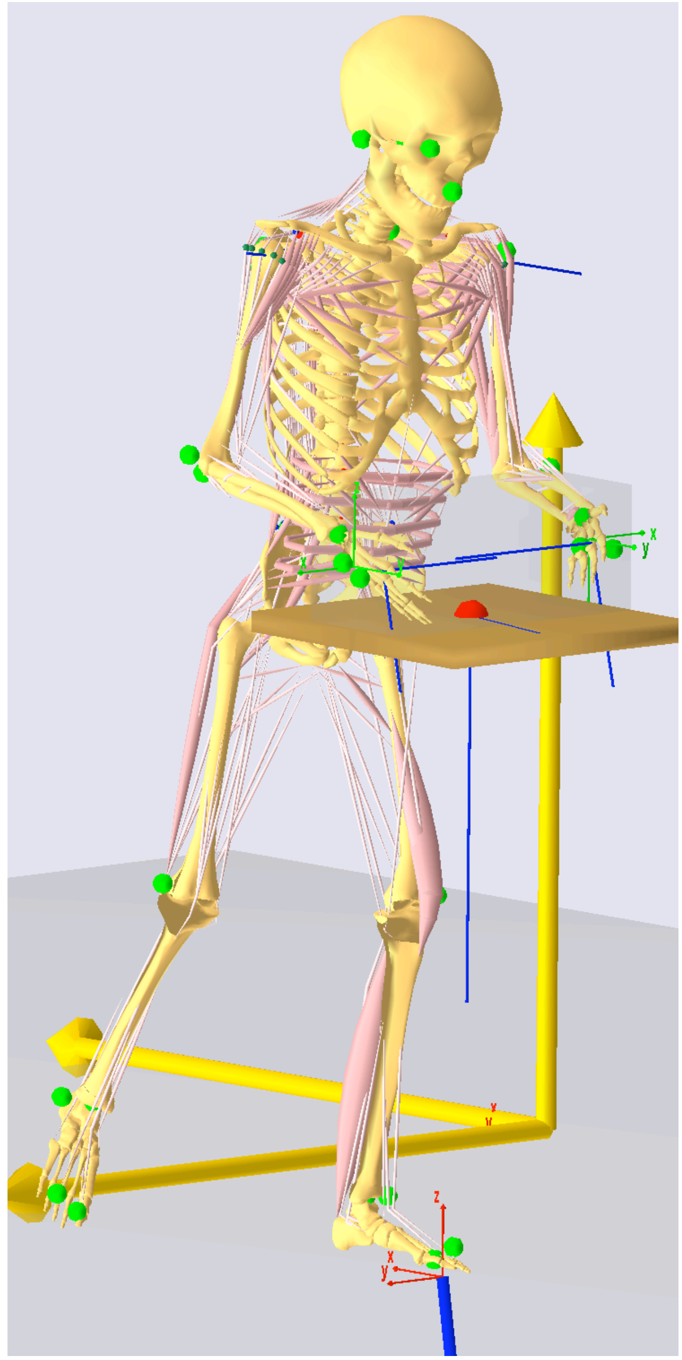

Figure 3. AnyBody model at load pickup.

RESULTS

The lower extremities of the participant were in a split stance posture during load pickup (Figure 3). The contralateral foot (defined here by the direction of turn) is supporting the full body weight of the participant while the ipsilateral limb is extended at the hip and in the process of externally rotating in preparation for the next step. The center of pressure (COP) is located at a point in the anterior and medial direction from a point bisecting the line defined by connecting the first and fifth metatarsal joints of the contralateral foot. The torso and neck are slightly flexed. Both arms are flexed such that the included angle at the elbow joint is approximately 90 degrees.

The moments at the low back (L5-Sacrum joint) for the dynamic and quasi-static simulations are presented in Figure 4. The moments are defined as positive for torso extension, lateral bending toward the right, and counterclockwise torsion of the torso. Peak low back moments were calculated for each direction and simulation type (Table 4). The quasi-static simulation results underestimated the dynamic simulation of the low back moments for the maximum values in all directions. The maximum flexion, lateral bending, and torsion moments for the static analyses were underestimated by 40.7%, 56.5%, and 47.2% respectively. The maximum flexion and lateral bending moments for the dynamic analysis occurred 0.24 s and 0.02 s prior to the corresponding peak moments obtained from the quasi-static analysis respectively. The maximum torsion moment for the dynamic analysis occurred 0.82 s following the corresponding quasi-static maximum.

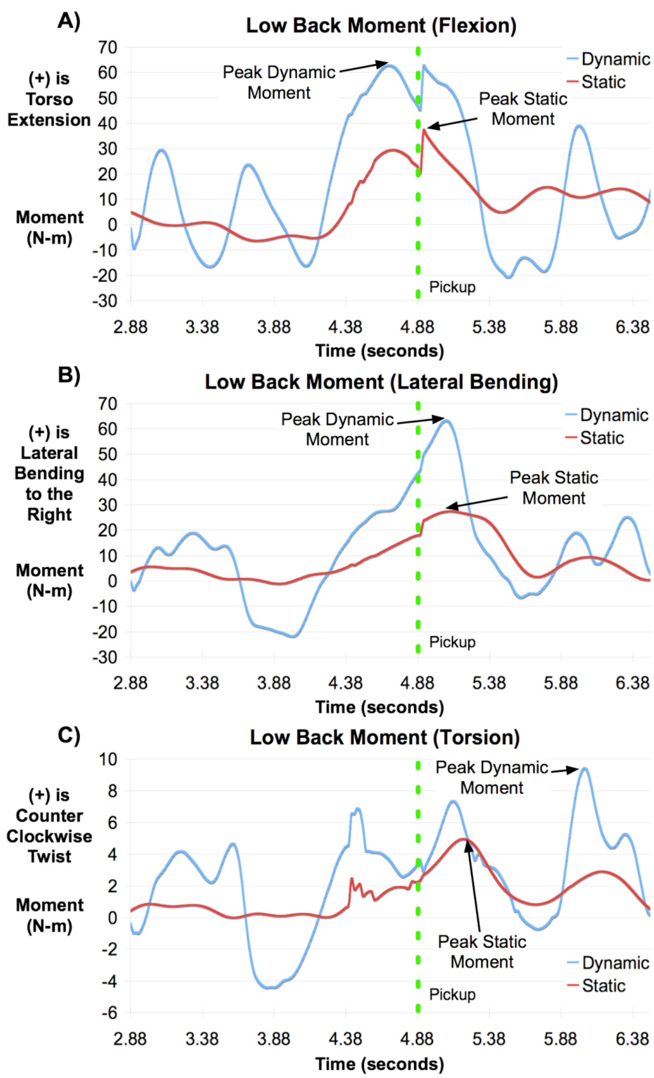

Figure 4. Low back moments profiles for A) flexion, B) lateral bending, and C) torsion for the dynamic and quasi-static simulations.

Table 4. Peak low back moments for the dynamic and quasi-static analyses.

	Dynamic Low Back Moment (N-m)			Quasi-Static Low Back Moment (N-m)		
Time (s)	Flexion	Lateral Bending	Torsion	Flexion	Lateral Bending	Torsion
4.68	**62.9**	27.8	3.7	29.3	12.9	1.8
4.92	62.7	49.7	3.0	**37.3**	23.7	2.7
5.08	54.9	**63.3**	6.7	25.5	27.5	4.2
5.10	53.8	62.7	7.1	24.3	**27.5**	4.4
5.20	41.7	41.2	6.0	18.6	26.8	**5.0**
6.02	38.7	18.1	**9.1**	10.9	9.2	2.4

*__Bold__ values denote maximum magnitudes recorded over the dynamic or quasi-static simulation for that direction.

The compression and shear forces at the low back (L5-Sacrum joint) are presented in Figure 5. The forces are truncated corresponding to the time the participant was completely supported by the force plate(s). The static analyses underestimated the peak dynamic compression and shear forces by 36.7% and 30.5% respectively. The peak and corresponding compression and shear forces for the dynamic and static analyses are presented in Table 5. The peak compression force for both simulations occurred 0.04 seconds after the pickup time. The maximum low back shear force for the dynamic simulation occurred at the same time as the peak compression force. However, the maximum shear force for the static simulation occurred 0.84 seconds prior to the dynamic result corresponding to the heel strike of the lead foot used during the load lift.

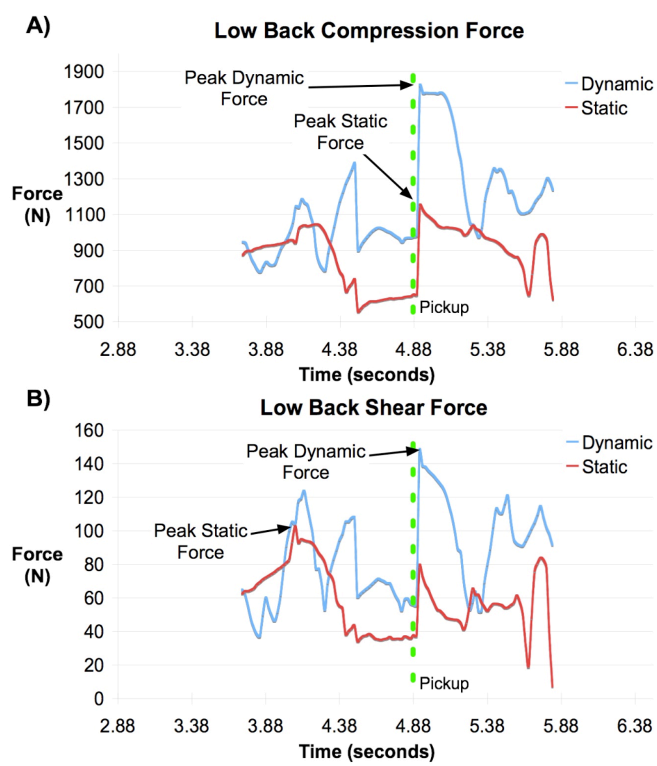

Figure 5. Low Back A) Compression and B) Shear forces for the dynamic and quasi-static analyses.

Table 5. Peak low back forces for the dynamic and quasi-static analyses.

	Dynamic Low Back Forces (N)		Quasi-Static Low Back Forces (N)	
Time (s)	Compression	Shear	Compression	Shear
4.08	1145.4	104.3	943.3	**102.9**
4.92	**1822.4**	**147.9**	**1152.8**	79.31

***Bold** values denote maximum magnitudes recorded over the dynamic or quasi-static simulation for that direction.

DISCUSSION

Dynamic Effects — This paper reports a substantial underestimate of internal body stresses when using quasi-static rather than a dynamic analysis of a non-stationary standing lifting task. Although both the quasi-static and dynamic estimates are subject to errors, the dynamic estimates are believed to be more accurate because inertial effects are included. The task was self-paced and the lifting tactics were self-selected, so the motion was free of some artificial constraints that have limited previous investigations of dynamic effects (see Hooper et al. 1998)

The peak dynamic flexion moment calculated in this study was underestimated by the static analyses by 40.7%. McGill and Norman (1985) reported a peak flexion moment underestimation of 16% when using quasi-static rather than dynamic methods to analyzing lifting tasks. Plamondon et al. (1995) reported a maximum deviation of 9 N-m for the low back extensor moment between a quasi-static and dynamic approach. This translates that for the reported maximum extension moment in the 'free-mode' lifting trial of 217 N-m, the peak moment would be underestimated in the static analysis by a maximum of 4.1%. However, in these studies, the loads lifted were significantly heavier than the one presented here; 11.6 kg for Plamondon et al. and 18 to 38 kg for McGill and Norman. The heavier loads may have caused the lifting velocities to be significantly slower than those observed in this study, resulting in a smaller contribution of dynamics and a smaller discrepancy between the quasi-static and dynamic results. In a study that used a similar mass (5.1 kg) to the one presented here (Tsuang et al. 1992) the average peak moments for a normal speed lift from the floor to waist level were underestimated by 34.5% with a quasi-static analysis.

The peak torsion moment calculated during the asymmetric lift was considerably less than those presented in the literature. The peak torsion moment for the dynamic analysis was 9.4 N-m and occurred 1.2 seconds following the pickup. A summary of peak twisting moments for six asymmetric lifting studies presented by Hooper et al. (1998) listed values ranging from 29 N-m to 100 N-m. One possible explanation for the discrepancy may be attributed to the difference in position (for example, lifting from the floor) and weight (up to 9 times heavier) in each of these studies. Unlike many previous studies, the feet of the participant in the current study were unconstrained, as in typical industrial tasks. However, the current analysis used data from a single subject and trial. Additional research should examine the repeatability across multiple lifts and participants and the relationship between unconstrained step placements and the self-imposed reduction or limiting of the magnitude of the torsion moment during lifting.

Low back compression and shear were also underestimated by the static analysis. The low-back compression and shear analyses are limited by the muscle-recruitment model employed by the AnyBody system. For these analyses, the optimization algorithm is based on a minimum fatigue criterion (Rasmussen and Christensen 2005). Use of a different criterion would have affected muscle activation patterns and hence the compression force obtained, but the comparison between the quasi-static and dynamic cases would have been similar.

Although both analyses for the data presented here yield the same conclusion relative to the safe lifting criterion set by NIOSH (< 3400 N compression force), the implication of the current analysis is that quasi-static analyses may fail to identify some jobs that exceed the criterion. With continued improvement in computing power and the advancement of software systems such as AnyBody, it is now feasible for dynamic analysis to be used in analyzing tasks in which body movement indicates a possibility for important dynamic effects.

Application of the AnyBody System — Kinematic definition of the AnyBody model was a significant impediment toward utilizing the AnyBody system as a "turn-key" dynamic solution. The AnyBody system (v2.0) requires that each model degree of freedom be uniquely defined. Kinematic redundancy in the data used to drive the model is not allowed, requiring the user to manually define how each degree of freedom is driven from the input dataset. Predefined repository functions for anthropometric scaling of the model geometry were limited to seven segment lengths and one width input. More advanced scaling techniques that utilized whole body mass and fat content are available; however these functions did not guarantee segment scaling consistent with measured values not included as part of the eight geometric inputs. Utilization of a published standardized set of anthropometric parameters for whole body scaling would improve the overall model generality and accuracy. A method for automatically handling the additional degrees of freedom usually available in the input data would implicitly allow for an arbitrarily defined set of whole body data (simulated or motion-captured) to drive the AnyBody model. Improvements in the anthropometric scaling and kinematic drivers would significantly reduce the time necessary to formulate a working model and shift the majority of time invested away from model definition to the primary purpose of task analysis.

As mentioned previously, the AnyBody Modeling System has the advantage of being able to incorporate dynamics in the analysis. However, the basic setup of this tool is much more general than the typical digital manikin tool and the models have a much higher level of detail with the inclusion of individual muscles. This, in combination with the additional complexity of the problem definition due to the dynamics, means that the setup and analysis of the problem is a time-consuming process requiring significant skill by the user. Current efforts on the system development side address these issues by the following means:

- *Motion-Capture Integration* — Driving a musculoskeletal model by means of motion-captured data is complex and currently requires considerable effort. The collected data is usually kinematically incompatible with the model linkage and contaminated with skin artifacts and other types of noise. In the present model these problems have been solved by careful data processing with much manual intervention, but this is not acceptable for routine ergonomic analyses. A current research project is addressing this problem and will generalize the kinematics module of the AnyBody Modeling System to accept over-determinate kinematics data and use the redundant information to identify and suppress noise.
- *User Interface* —The current system setup allows for musculoskeletal modeling of any living organism or machine. This generality also affects the interface and complicates it beyond what is necessary to work with human models. Future system developments will facilitate the development of applications based on a fixed model (usually a human) for which the user interface can be much simpler and more applicable in a clinical situation.
- *Model Completion and Validation* —The repository of human models for the AnyBody system is still missing a few vital parts of the human body: the hands, the feet, the thoracic and cervical spine segments. In addition, the models have been validated for relatively few postures and loading situations. Completion and validation is an ongoing and probably open-ended process.

Dynamic Analysis for Ergonomics — The limited analysis presented in this paper demonstrated both the feasibility and utility of inverse dynamics for ergonomic applications. The model estimates of low-back stresses, in particular, were substantially higher when considering inertial effects. Considerably more work is required to bring dynamic analysis into routine use for ergonomic analysis. Several of the important issues relating to model scaling, body segment parameters, and motion-capture integration are noted above. One of the biggest challenges, however, is to free the dynamics analysis from being driven by motion capture data. Currently, motion capture is the only way to generate quantitatively realistic motions of sufficient quality for inverse-dynamics analysis. Extensive research is underway to develop improved human motion simulation algorithms that may be suitable for dynamic analysis. For example, the HUMOSIM Framework is a modular motion-simulation structure the predicts the kinematics of motion (Reed et al. 2006). One intriguing possibility is the use of forward dynamics, in which the motions are predicted by muscle-level control, using more general kinematic-level models, such as the HUMOSIM Framework, as training input to the forward dynamics models (e.g., see Rasmussen et al. 2006). If such methods can be validated, dynamic analysis will be available for a much wider range of tasks.

ACKNOWLEDGMENTS

This research was sponsored by the partners of the Human Motion Simulation (HUMOSIM) program at the University of Michigan. HUMOSIM partners include DaimlerChrysler, Ford, General Motors, International Truck and Engine, United States Postal Service, U.S. Army Research and Development Engineering Command (RDECOM).

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ADDITIONAL SOURCES

Additional information on the research presented in this paper is available from the Human Motion Simulation Laboratory (http://www.humosim.org/) and from AnyBody Technology (http://www.anybodytech.com/).