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Investigation of the mass distribution of a detailed seated male finite element model

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Running Title: Mass Distribution of Seated Male FEM
Abstract

Accurate mass distribution in computational human body models is essential for proper kinematic and kinetic simulations. The purpose of this study was to investigate the mass distribution of a 50th percentile male (M50) full body finite element model (FEM) in the seated position. The FEM was partitioned into 10 segments, using segment planes constructed from bony landmarks per the methods described in previous research studies. Body segment masses and centers of gravity (CGs) of the FEM were compared to values found from these studies, which unlike the present work assumed homogeneous body density. Segment masses compared well to literature while CGs showed an average deviation of 6.0% to 7.0% when normalized by regional characteristic lengths. The discrete mass distribution of the FEM appears to affect the mass and CGs of some segments, particularly those with low density soft tissues. The locations of the segment CGs are provided in local coordinate systems, thus facilitating comparison to other full body FEMs and human surrogates. The model provides insights into the effects of inhomogeneous mass on the location of body segment CGs.

Keywords: human body model, center of gravity, mass distribution, model validation

Word Count: 2,385
Introduction

Simulations of blunt trauma are now possible due to advances in computational human body modeling via finite element analysis. Finite element analysis is often used due to its capability of handling complex geometries and modeling the non-linear behavior of tissues. This study focuses on a full body finite element model (FEM) of a 50th percentile male (M50 model) that is in development as part of the Global Human Body Models Consortium (GHBMC) project.

FEMs of specific body parts and regions have been developed to analyze specific injuries. Full body FEMs offer the advantage of observing injuries that may occur as the whole body dynamically interacts with its environment. Many verification practices are needed to ensure a model’s biofidelity including correct anthropometry, geometry, material properties and boundary conditions in addition to numerical validation vs. literature studies of kinematics and kinetics. During a dynamic event, model response will also be a function of the mass distribution, i.e., the location of the centers of gravity (CGs) of its various components. Due to this, it is imperative also to characterize the CGs of full body FEMs used in dynamic modeling.

Therefore, the focus of this work is on the mass distribution of the GHBMC seated 50th percentile male full body FEM. The objective of this research is twofold. The first aim is to demonstrate a technique to investigate the mass distribution of the model using methods described in the literature. The second is to compare the masses and CGs of the M50 model to these two studies.
Methods

The geometry of the M50 model is based on medical images and surface scans of a subject who closely matched standard definitions for the midsized male. Detailed descriptions of model development are outside the scope of this paper, but can be found in the literature. The M50 model (v3.5) consists of 1.9 million elements, each with an assigned density (Figure 1). Since density values can be readily found for many body components the model captures the variations in mass distributions.

A number of studies to determine segmental CGs of the human body have been conducted. The two studies chosen as a benchmark for the current work were by McConville et al. (1980) and Robbins (1983). McConville’s study was selected because body segments CGs were provided with respect to a three dimensional local coordinate system defined within the segment, although the subjects were in a standing posture. Robbins adapted this work for the seated posture by adjusting the body segments and local coordinate systems. Since the locations of bony landmarks on M50 were known, these studies were deemed the best for comparison. A brief description of both study methodologies follows. In each study, the full body was divided into 10 segments using cutting planes described by bony landmarks. CGs were determined for each of the 10 body segments using a homogenous density assumption. The CGs were then defined in local coordinate systems which were also created from bony landmarks. Though the homogenous density assumption was made to simplify calculations, both studies acknowledged it as a limitation.

The methods in this study focus on four main goals. The first was to develop 10 section planes from the literature descriptions, based on bony landmarks. The second was to use these planes to partition the full body FEM into segments. The masses and CGs of each segment were then determined via FEA pre-processing software (LS-PrePost v. 3.2, Livermore Software Technology Corp, Livermore, CA). Third, CGs of each region were determined using methods
described by McConville et al.\textsuperscript{27} and Robbins\textsuperscript{28}. Finally, the CG locations from these literature studies were compared to the CG locations from M50 by calculating the linear distance between each, and normalizing those by a set of characteristic lengths. To ensure that data from McConville et al. and Robbins was appropriate for this study, height and weight of the subjects were compared in the referenced study were compared to the individual used in the model development of M50.

Planes were used to divide M50 into the following 10 segments: head, neck, thorax, abdomen, pelvis, upper arm, forearm and hand, thigh, calf, and foot. To accommodate M50’s seated posture, segment planes defined by McConville et al.\textsuperscript{27}, were modified for the abdomen, hip, knee and ankle planes per the descriptions of Robbins\textsuperscript{28}. Two alternate sectional planes were used to account for the soft tissue in the abdominal region caused by the seated posture. Since knees and ankles of M50 were flexed, the knee and ankle section planes were rotated 58° clockwise about the Y (Left-Right) axis, matching the toe pan angle of the model. Given that the exterior of M50 is sagittally symmetric, partitioning the extremities was performed only along the right side. Figure 2 shows the planes used to partition FEM into segments.

The mass, and CGs of each segment were calculated (LS-PrePost v. 3.2) for each isolated body segment of the FEM. To determine CG values based from literature data, a local coordinate system (LCS) was established per body region from the bony landmarks on the M50 model. Three mutually orthogonal planes were constructed, following the methods of McConville et al. and Robbins. The origin of the LCS occurred at the intersection point of the three planes. For each body region, McConville et al. provided the Cartesian trajectory in the LCS from the region’s CG to various bony landmarks. Working backwards from each bony landmark (taking specific nodes to represent landmark locations) and using these trajectories, a
CG location was calculated. Multiple landmarks were used and averaged to represent the McConville CG (MCG) for that region. Due to the large number of degrees of freedom and variation of curvature that occurs in the neck region, the CG of the neck region was located from only one landmark, the spinous process of the 7th cervical vertebrae.

CGs reported by Robbins (RCGs) were given with respect to the McConville LCSs for all regions except the thorax and abdomen. For these two regions, LCSs specific to Robbins’ work were constructed. Figure 3 shows an example of the LCS and the left anterior superior iliac spine (ASIS) landmarks used to find a CG for the pelvis. Here, the local axes are X’, Y’, and Z’, and the distance in each direction is given as ΔX’, ΔY’, ΔZ’. In this example, the left ASIS landmark (blue pentagon) was used to find the pelvis center of gravity (red cube) following Cartesian trajectories (ΔX’, ΔY’, ΔZ’) provided in the literature study. The average CG (n = 4) is the yellow sphere.

The Euclidean distances between each regional CG were determined and normalized by a characteristic length found for each region. Since the geometry of each segment varied, the characteristic length was set as the longest length of a 6 sided bounding box enclosing the segment (Error! Reference source not found.). Due to differences in posture between M50 and the McConville data set, distances between FEM CGs in the thorax and abdomen regions are only presented vs. Robbins data.

Results

Good agreement with the masses from the literature studies were found for all body segments. All 10 body segments were within the range of masses given in McConville. However, the Thorax and Abdomen were both at the extremes of these ranges; the Thorax in M50 was low in the range and the Abdomen was high in the range (Error! Reference source
not found.). In regards to full body measurements, the standing height of M50 was within 1.3% of the mean of both McConville and Robbins’ studies and the weight was within 3.4%.

CG locations were also determined for the FEM and are given in Error! Reference source not found.. Comparisons to the experimental values were made by normalizing the differences by the characteristic length of each segment (Error! Reference source not found.). These comparisons showed absolute distances ranging from a minimum of 10.5 mm in the Thorax to a maximum of 44.8 mm in the pelvis (both compared to Robbins). These resulted in percent differences ranging from 2.6% to 14.0%, respectively. The normalized differences seen between McConville and Robbins ranged from 0.0% to 12.6%. The location of the CGs for all body regions are shown in addition to the local coordinate systems for each segment (Figure 4). Since mid-sagittal symmetry was enforced in M50, error along the Y axis of the LCS was less than 1% of the characteristic length in all axial CGs. Note that McConville’s CGs for the thorax and abdomen are included in these plots for completeness.

Discussion

A method to determine segmental masses and CG values for a full body FE model of the seated male was presented. The method was adapted to an FE model from the studies of McConville et al. and Robbins. Comparisons to the results of those studies were made. In the current study, no assumptions were made about the morphology and location of internal structures as these were discretely modeled in the M50. In the studies used for comparison, a homogenous density was assumed. Landmark locations for M50 can be found in the literature.31 In general, the FEM based values were similar to the established literature studies, but highlighted some differences. These differences are within the range of differences found between the McConville and Robbins studies, seen in Table 1, and were deemed acceptable.
The method described by McConville and Robbins provides a robust technique for determining segmental mass distribution properties. An advantage of this method is the use of bony landmarks to describe the cutting planes, local coordinate systems, and CG locations. This allows for the application of the method to a variety of body types and postures. Robbins made adjustments to the methods of McConville to correct for a seated posture. These corrections involved modifying a bony landmark used for the thorax and abdomen. In McConville one of the bony landmarks was the level of the 10th rib at the mid-spine. This does not translate to the same point with a seated subject so Robbins created a new landmark which he called R10/Back Perpendicular. This point is made by constructing a line from the 10th rib landmark that is perpendicular to the line between the T12 and L5 vertebrae. By adjusting for the seat back angle, this new landmark provides a similar location as the McConville landmark.

The mass distribution of FE models are generally not explicitly studied, yet mass properties make significant contributions to the overall kinematics of the model. Blunt impacts can be thought of as including both gross rigid-body movement (i.e. flailing of the upper extremity) in addition to local deformations (i.e. chest wall loading a seatbelt or airbag). Thus, it is necessary to have accurate segmental CGs in addition to appropriate material models to capture both types of displacement. A method was described in this paper which can be applied to human body FE models to ensure biofidelic CG locations and mass properties. Given the data provided and established methods used to acquire it, the CG locations of the M50 model could be used as a benchmark for future model development or even physical surrogate development.

There are two limitations that merit further discussion. The first limitation involves the nature and number of anatomical structures modeled in M50. Not every structure in the human body is represented in M50, rather, the anatomy focused on what was required to predict
commonly encountered injuries in blunt trauma. This has ramifications at both the macro and micro scales. At the macro scale, for instance, the spinal cord, mesentery, and blood vessels less than 4 mm in diameter are examples of structures that were not explicitly modeled. These anatomical components were excluded because they do not represent common injuries in car crashes and, in the case of the mesentery and vessels, are highly variable between subjects. In regards to potential limitations that result from these specific modeling considerations, the spinal cord, with an average mass of 30 g, is thought to not be massive enough to affect the CG of the regions in which it is located. While the mesentery is not explicitly modeled, the bowel is represented with shell elements of the appropriate density (8.44x10^{-6} kg/mm^3) and a control volume meant to model the presence of gas within it. As part of the design of the model, only vessels of greater than 4 mm were explicitly modeled. It is assumed that smaller vessels and the microvasculature would not significantly affect the CG calculations. Density variation can only be captured on the scale of the average element volume, 34.8 mm^3, which translates to an average element edge length of 3.25 mm. Therefore, on the micro scale, all anatomical features that are smaller than the typical element size were not explicitly modeled. Yet, the constitutive equations describing the material behavior account for the biomechanical response determined by structures on this scale.

Despite the above modeling considerations, the model is quite anatomically detailed and includes over 400 individual components. After summing the mass of the nearly 2 million elements of M50, the total is comparable to sample average from two studies used for comparison, and is nearly the exact weight of the subject used as the basis for model construction. The mass of the M50 model, and the means of the McConville and Robbins studies were 78.6, 76.6, and 75.2 kg, respectively. Densities were not tuned to obtain agreement
in the total mass so it can be inferred that the absent anatomy does not substantially contribute to the mass of the body.

A second limitation is that there is no experimental data in the literature that allows for a more direct comparison to M50 since no datasets were available that assumed a heterogeneous density while also providing segment based LCSs. Though there was generally good correlation between most body segment masses and CGs of M50 with the literature, there were some areas of variation. These areas include the thorax, and pelvis. Since posture affects were accounted for, the differences found in this study are thought to be a function of the use of a discrete and heterogeneous representation of density in the human body. More precise experimental data would be required to validate or reject this hypothesis.

Overall, the results of this study contribute to the development of the M50 model. Accurate human body models will continue to enhance the understanding of crash injury mechanisms, and will ultimately reduce injuries and fatalities for individuals sustaining blunt injury.
Acknowledgement

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Figure 1. Finite element model of M50. The exterior skin has been removed on the right side to demonstrate the internal body structures included.
Figure 2. Segment planes on the finite element model of M50. Anterior and Sagittal views of the segment planes are shown.
Figure 3. Pelvis center of gravity from the left anterior superior iliac spine (ASIS) landmark. Isometric, sagittal, anterior, and coronal views are also shown with the local coordinate system. The ASIS landmark is the pentagon, the CG from the ASIS is the cube, and the average CG (n = 4) is the sphere.
Figure 4. Comparison plot of body region CGs between the M50 finite element model, McConville et al. and Robbins. Local coordinate systems are shown (left) separately.
Table 1. Mass distribution comparison between FEM and McConville and Robbins. Segment masses of the FEM are compared to the range presented by McConville et al. Relative distances between FEM CGs to McConville and Robbins CGs, normalized by characteristic lengths, are also shown with relative distances between McConville and Robbins.

<table>
<thead>
<tr>
<th>Region</th>
<th>Segment Masses</th>
<th>FEM CG Location</th>
<th>FEM Relative to MCG</th>
<th>FEM Relative to RCG</th>
<th>MCG Relative to RCG</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>FEM (kg)</td>
<td>McConville Range</td>
<td>X (mm)</td>
<td>Y (mm)</td>
</tr>
<tr>
<td>Head¹</td>
<td>4.51</td>
<td>3.54 – 4.73</td>
<td>3</td>
<td>0</td>
<td>31</td>
</tr>
<tr>
<td>Neck¹</td>
<td>1.01</td>
<td>0.70 – 1.50</td>
<td>45</td>
<td>0</td>
<td>74</td>
</tr>
<tr>
<td>Thorax¹²</td>
<td>17.75</td>
<td>17.88 – 38.39</td>
<td>76</td>
<td>1</td>
<td>163</td>
</tr>
<tr>
<td>Abdomen²</td>
<td>3.76</td>
<td>0.91 – 3.87</td>
<td>0</td>
<td>-1</td>
<td>-36</td>
</tr>
<tr>
<td>Pelvis¹</td>
<td>9.78</td>
<td>7.08 – 16.24</td>
<td>-79</td>
<td>0</td>
<td>-34</td>
</tr>
<tr>
<td>Upper Arm¹³</td>
<td>2.71</td>
<td>1.18 – 2.94</td>
<td>2</td>
<td>19</td>
<td>-177</td>
</tr>
<tr>
<td>Forearm &amp; Hand¹³</td>
<td>1.50</td>
<td>1.28 – 2.54</td>
<td>-7</td>
<td>21</td>
<td>-145</td>
</tr>
<tr>
<td>Thigh¹³</td>
<td>10.91</td>
<td>7.01 – 12.96</td>
<td>5</td>
<td>39</td>
<td>-186</td>
</tr>
<tr>
<td>Calf¹³</td>
<td>4.01</td>
<td>2.63 – 4.78</td>
<td>0</td>
<td>-43</td>
<td>-160</td>
</tr>
<tr>
<td>Foot¹³</td>
<td>1.32</td>
<td>0.68 – 1.24</td>
<td>-89</td>
<td>-7</td>
<td>9</td>
</tr>
</tbody>
</table>

¹References local coordinate systems described by McConville for CG Locations
²References local coordinate systems described by Robbins for CG Locations
³References right side only. To find corresponding left side value, use the negative of the Y component while leaving the X and Z values same.
⁴Calculated from densities and volumes reported by McConville
⁵Normalized by characteristic lengths