Simulift: A Simulation Model of Human Trunk Motion

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In this paper, the authors present a deterministic simulation model, which they call Simulift, of trunk-muscle activity and intra-abdominal pressure during a sagittally symmetric trunk exertion. Simulift is a descriptive model that quantifies the time-varying loading of the spine based on observed internal forces. Recent findings about the time sequence of events during trunk motion and established equilibrium formulas provide the theoretical bases for the simulation. A profile of electromyographic activity in ten trunk muscles and intra-abdominal pressure is updated as simulated time passes, or as the trunk motion is simulated. Input to the model includes a list of motion-event times and a set of profile-component-behavior data. Simulift is an "impulse" model that computes instantaneous and time-integrated statistics on individual muscle activity, intra-abdominal pressure, compression, lateral shear, and anterior shear as the profile components of the simulated subject change. Computer results for the simulation model are presented. [Key words: spinal model, internal forces, trunk motion, trunk musculature]

THE SIGNIFICANCE of the moments imposed about the spine and their relation to loading on the spine have been demonstrated by researchers such as Chaffin and Baker[2] and Schultz and Andersson.[6] A major portion of the loading on the spine during a trunk motion has been shown to be attributable to the internal forces within the trunk, since these forces create countermoments to the moments generated external to the body during trunk motion.[6] These countermoments are generated in close proximity to the spine (fulcrum); hence, they produce very large forces.

In their paper, Schultz and Andersson proposed a cross-sectional equilibrium model designed to summarize the relationships between the internal forces, the internal moments, and the compressive and shear forces on the spine under static conditions. Their model consisted of six equations, describing the forces and moments about the spine, in 14 unknowns (10 muscle forces, intra-abdominal pressure [IAP], compression, anterior shear, and right-lateral shear); hence, its solution is statically indeterminate. Attempts were made to arrive at solutions to this model by fixing some of the unknowns, for example, by assuming antagonistic activity and IAP were 0. This approach leads to solutions in which few of the unknowns have non-zero values. Schultz et al[7-9] have validated the model suggested by Schultz and Andersson under static conditions.

Marras et al[3] found that, under maximal dynamic and maximal static exertions, all ten muscles included in Schultz and Andersson's model, as well as IAP, are always active to some degree, and this is especially true for dynamic exertions. Marras et al also found that the internal forces change substantially and rapidly during a dynamic exertion. The integrated electromyographic (EMG) recordings of this activity show the spontaneous effects of changes in muscle activity and IAP under dynamic conditions. This work points out the need for models that can evaluate instantaneous spine loading.

Recently, Marras and Reilly[4] have conducted an investigation into the patterns of trunk-muscle recruitment and activation during sagittally symmetric trunk exertions. They have summarized these recruitment patterns under maximal static exertions and maximal dynamic (isokinetic or zero acceleration) exertions observed at different (constant) velocities, and have identified key motion events that occur during trunk exertions. This study of trunk-muscle-recruitment patterns has made possible new modeling approaches for spine-loading models, particularly models that focus on the changes in the loads on the spine throughout trunk motion. These new approaches permit the evaluation of instantaneous or "impulse" loading on the spine, as well as the quantification of loading variability.

In this article, we present a descriptive deterministic simulation model of internal spine loading, called Simulift, into which we have incorporated the time sequence of motion events. This "impulse" model of the internal trunk forces can be used to characterize, quantify, and analyze the loading of the spine under static and controlled dynamic exertions. Of particular interest is the characterization of spine loading as a function of trunk velocity. We have employed Simulift with data collected under five experimental conditions, a static condition and four dynamic (isokinetic) conditions. In developing Simulift, we have capitalized on the equilibrium model of Schultz and Andersson, by embedding it in our simulation, and the trunk-muscle-recruitment-pattern study by Marras and Reilly.

Simulift can measure spine loading in relative terms or in absolute terms, given appropriate anthropometric assumptions. We have chosen to measure spine loading in relative terms here because our objective is to characterize spine loading as a function of trunk velocity. This approach permits us to characterize spine loading over a broad range of anthropometric parameters. Differences in spine loading between experimental conditions are not affected by the use of relative, rather than absolute, forces. For example, we do not estimate the actual magnitude of the compressive forces on the spine, but we are able to estimate when the compressive forces are greatest during an exertion and what the relative change in peak compression is from one condition to another.

DATA FOR MODEL PARAMETERS

The data used to parameterize Simulift are the same data used by Marras and Reilly[4] in their investigation into the patterns of trunk-muscle recruitment during sagittally symmetric trunk exertions. Readers are directed to their paper or to Marras et al[3] for a description of how the data were collected.

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Electromyographic data on each of ten trunk muscles (the latissimus dorsi left [LATL] and right [LATR], the erector spinae left [ERSL] and right [ERSR], the external oblique left [EXOL] and right [EXOR], the internal oblique left [INOL] and right [INOR], and the rectus abdominus left [RCAL] and right [RCAR]) and IAP readings were collected under five experimental (lifting) conditions:

1) Isometric exertion; trunk bent forward at 67.5° (from lumbosacral junction of the lumbar spine);
2) Isokinetic exertion; trunk bent forward at 67.5° at outset; trunk in erect posture at finish; 25% of subject's maximal trunk velocity.
3) Same as 2), except 50% of subject's maximal trunk velocity.
4) Same as 2), except 75% of subject's maximal trunk velocity.
5) Same as 2), except 100% of subject's maximal trunk velocity.

Once collected, the data on magnitudes of EMG activity and IAP were normalized over the experimental conditions for each subject and each muscle or IAP. This was necessary to compare forces between muscles and subjects, since EMGs (and IAP) are unique for each muscle and subject. For any muscle, the normalized-data values indicate the percentage of a subject's maximum (observed) EMG activity that the muscle is exhibiting at the associated time. A normalized-data value for IAP is interpreted similarly.

The data were normalized with respect to time as well. For each subject and each experimental condition, the activity-time axis was normalized to the interval [0,1]. Following this second data normalization, the normalized magnitude data for different subjects or for a single subject under different experimental conditions could be plotted against a common time axis and compared in a meaningful way. This is precisely what Marras and Reilly did in their analysis of the patterns of trunk-muscle recruitment.

Plots of the response variables (normalized EMG and IAP magnitudes) versus normalized time showed that for nearly every response variable-experimental condition-subject combination, response variable i (i = 1[LATL], 2[LATR], . . . , 9[RCAL], 10[RCAR], 11[IAP]) increased almost monotonically from a resting level f_{i1} once the response's activity began at time t_{i1} (f_{i1} is the value of response variable i at normalized time t_{i1}), reached a peak level f_{i2} at time t_{i2}, and then decreased almost monotonically until it reached a terminal resting level f_{i3} at time t_{i3}. A typical plot of a response variable versus time is shown in Figure 1. This consistent pattern among the response variables permits us to represent the response as a triangular function of normalized time, as illustrated in Figure 1.

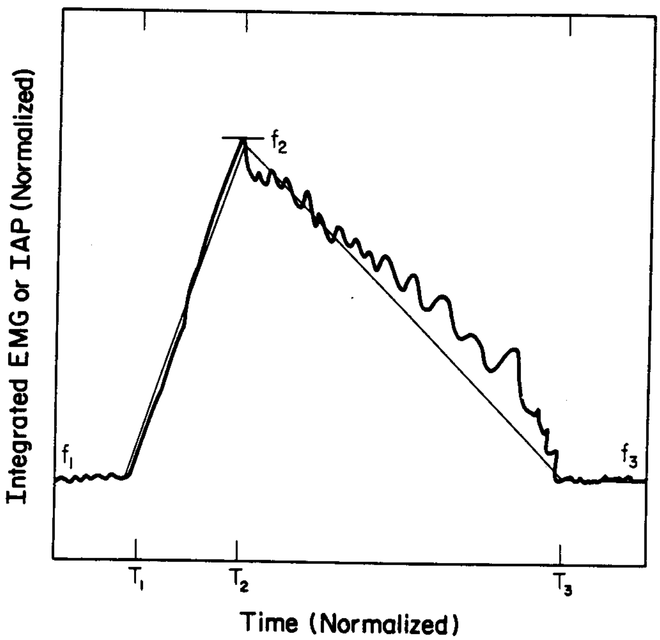

Fig 1. Typical plot of response variable vs. normalized time.

In the last section, we reported that the plots of the profile components versus time exhibited a consistent pattern, ie, initial resting level, monotonic increase, peak, monotonic decrease, terminal resting level. We assume that the monotonic increase in profile element i between the associated motion-event times t_{i1} and t_{i2} is actually a linear increase. We make an analogous assumption for the monotonic decrease in profile element i between the times t_{i2} and t_{i3}. Therefore, the value of profile component i can be estimated at any time, $0 \leq t \leq 1$, using the following formula:

$$F_i(t) = \begin{cases} f_{i1} & \text{if } 0 \leq t \leq t_{i1} \\ f_{i1} + (f_{i2} - f_{i1})(t - t_{i1})/(t_{i2} - t_{i1}) & \text{if } t_{i1} \leq t \leq t_{i2} \\ f_{i2} - (f_{i2} - f_{i3})(t - t_{i2})/(t_{i3} - t_{i2}) & \text{if } t_{i2} \leq t \leq t_{i3} \\ f_{i3} & \text{if } t_{i3} \leq t \leq 1 \end{cases} \quad (1)$$

At any point in simulated time, the values of the profile elements can be used in the equations of the equilibrium model of Schultz and Andersson[6] to calculate indices for compression, anterior shear, and (right-) lateral shear. Since this cross-sectional equilibrium analysis describes the physical orientation of the moments about the spine, it should be valid for evaluating dynamic (isokinetic) changes within the trunk as well as static forces. An assumption that is implicit in this approach is that the force generated by each trunk muscle is proportional to that muscle's EMG signal. (Such a relationship between EMG signals and forces was reported by Bigland and Lippold[1] for constant velocity contractions.)

The EMG data used in Simulift are measured in terms of the percentage of the maximal observed EMG activity in each muscle. Intra-abdominal pressure data are measured similarly. The equilibrium model of Schultz and Andersson assumes that all response variables are measured in Newtons. Hence, the index values for compression, anterior shear, and lateral shear that are calculated based on this model are relative measures for the actual forces. Though we do not calculate actual compression, for example, we are still able to determine when the compressive forces acting on the spine during a simulated trunk motion are greatest from the index for compression. We are interested in the relative change in the forces as a function of the velocity conditions, not with the absolute forces themselves.

DESCRIPTION OF THE SIMULATION MODEL

Simulift is an "impulse" model designed to simulate a human trunk profile during a static exertion or during a dynamic motion. The trunk profile consists of the EMG activity in ten trunk muscles (LATL, LATR, ERSL, ERSR, EXOL, EXOR, INOL, INOR, RCAL, and RCAR) and the level of IAP. These profile components represent the internal loads acting at the lumbosacral junction (L5-S1). This profile is evaluated at 35 critical times, at the beginning of the simulation (normalized time 0), at each of 33 motion-event times, and at the end of the simulation (normalized time 1). The 33 motion-event times are the times at which a profile component begins to increase from its initial resting level (t_1), the times at which a profile component reaches its zenith (t_2), and the times at which a profile component reaches its terminal resting level (t_3). There are three motion-event times for each of the 11 profile components.

This impulse model is descriptive. Given EMG and IAP data collected under static or isokinetic dynamic conditions, Simulift can describe the time-varying loading of the spine. Though Simulift cannot describe spine loading under conditions for which there are no data, the results observed with this model may ultimately lead to more sophisticated prescriptive models of spine loading, because Simulift characterizes how muscle actions change as a function of trunk velocity.

The equilibrium model of Schultz and Andersson is applied

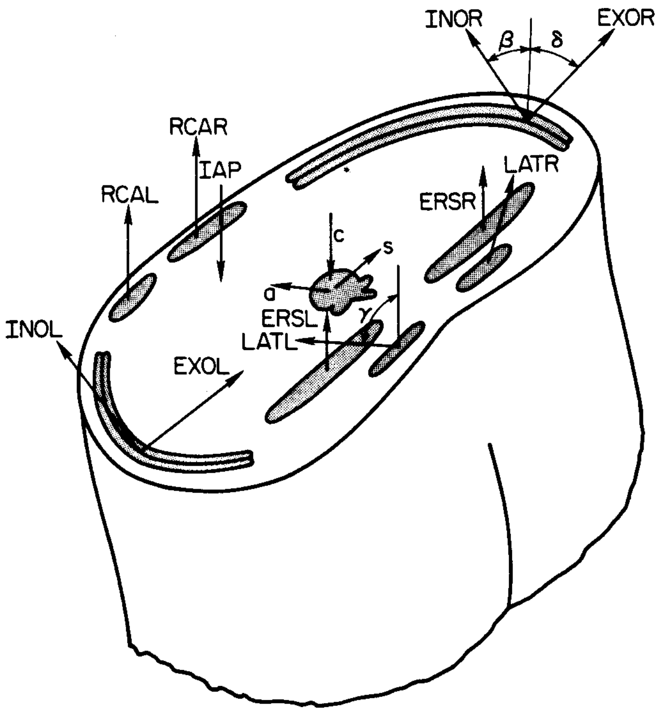

Fig 2. Cross-sectional view of human trunk at the lumbo-sacral junction.

directly in Simulift. However, rather than solving three equations in 14 unknowns (ignoring the 3 moment equations) under certain anthropometric assumptions, Simulift provides relative values for 11 of the (unknown) forces, thus permitting calculation of indices for compression, anterior shear, and lateral shear.

Figure 2 depicts a cross-sectional view of the trunk at the lumbosacral junction of the lumbar spine. Each of the forces of interest is shown in the figure, as are the angles of the latissimus dorsi (γ), the external oblique (δ), and internal oblique (β) muscles.

Before the simulation actually begins, a list of the motion-event times is specified, as are assumed values for the angles β, δ, and γ. The motion-event times are sorted and listed on a calendar of events in increasing order of event time.

The first event, the start-of-simulation event, occurs at simulated normalized time 0. At this time, the trunk profile is initialized. All profile components are set equal to their initial resting levels (percentage of maximum EMG or IAP). Indices for compression (c), anterior shear (a), and lateral shear (s) are calculated based on the current trunk profile, using the following equations derived from Schultz and Andersson's model:

$$c = ersl + ersr + rcal + rcar + \cos(\gamma)(latl + latr) + \cos(\delta)(exol + exor) + \cos(\beta)(inol + inor) - iap \tag{2}$$

$$a = \sin(\delta)(exol + exor) - \sin(\beta)(inol + inor) \tag{3}$$

$$s = \sin(\gamma)(latr - latl) \tag{4}$$

(The lower case letters that correspond to our notation for muscles and IAP represent simulated EMG and IAP values, respectively.) The simulated time clock is then advanced to the time of the first motion event.

At each of the motion-event times, the simulated human-trunk profile is updated according to Equation 1. Indices for compression, anterior shear, and lateral shear are computed using Equations 2, 3, and 4, respectively. The time clock is advanced, and the simulation continues. Finally, the end-of-simulation event occurs at simulated normalized time 1.

Time-integrated statistics on the indices and the profile components are calculated. It is assumed in the calculation of the summary statistics that the values of the indexes change linearly over time between motion-event times.

SUMMARY OF SIMULIFT RESULTS

We have encoded a computerized version of Simulift in Fortran. In order to simulate a lifting motion, three points, (t_{i1},f_{i1}), (t_{i2},f_{i2}), and (t_{i3},f_{i3}), must be specified for each profile element i. Assumed values for the angles γ, δ, and β are also specified. We assume that each of these angles is 45°, just as Schultz and Andersson do (p 80).[6] A 35-event simulation, including input and output time, is executed in about 0.30 CPU seconds on an IBM 3081-D at The Ohio State University.

The data input to Simulift can apply to an individual subject for any experimental condition. In our case, we averaged the event times and profile-component magnitudes of the ten subjects to arrive at an input data set for each experimental condition. (See Table 1 for a listing of the 33 points of the form $(t_{ik},f_{ik}$, $i = 1,2,\ldots,11$, $k = 1,2,3$, for the 50% velocity condition.) We ran Simulift for each of the experimental conditions. Our results are summarized below.

Figures 3 through 5 are graphical displays of the respective indices calculated during the Simulift run for each experimental

Table 1. Normalized Time and Profile Component Data (50% Velocity Condition)

Profile component	t_{i1}	f_{i1}	t_{i2}	f_{i2}	t_{i3}	f_{i3}
LATL	0.10700	0.18800	0.24100	0.74400	0.75900	0.18300
LATR	0.11300	0.21600	0.23889	0.77889	0.78900	0.25700
ERSL	0.15700	0.12600	0.35600	0.78400	0.73800	0.10300
ERSR	0.12800	0.18500	0.40000	0.71100	0.70700	0.18200
EXOL	0.13625	0.22375	0.35700	0.65500	0.84250	0.25125
EXOR	0.35200	0.41600	0.44125	0.73625	0.81667	0.48667
INOL	0.18300	0.30700	0.45250	0.79125	0.83556	0.35333
INOR	0.29300	0.27000	0.48800	0.68400	0.77778	0.29000
RCAL	0.29100	0.44600	0.33667	0.79222	0.64900	0.54400
RCAR	0.27000	0.11889	0.46833	0.59167	0.72000	0.14111
IAP	0.13750	0.07750	0.23500	0.79400	0.95400	0.19000

LATL = latissimus dorsi left; LATR = latissimus dorsi right; ERSL = erector spinae left; ERSR = erector spinae right; EXOL = external oblique left; EXOR = external oblique right; INOL = internal oblique left; INOR = internal oblique right; RCAL = rectus abdominus left; RCAR = rectus abdominus right; IAP = intra-abdominal pressure.

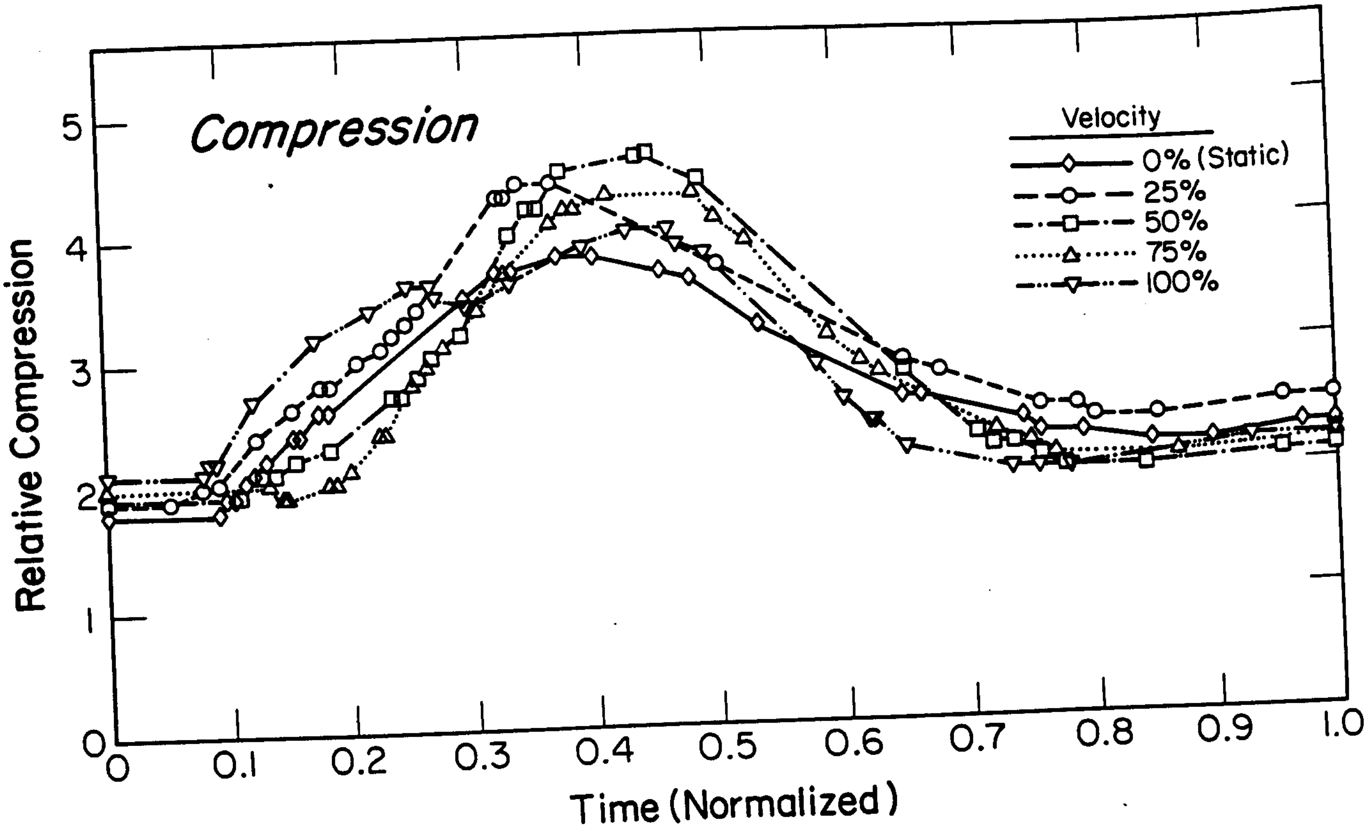

Fig 3. Plot of compression index over simulated normalized time.

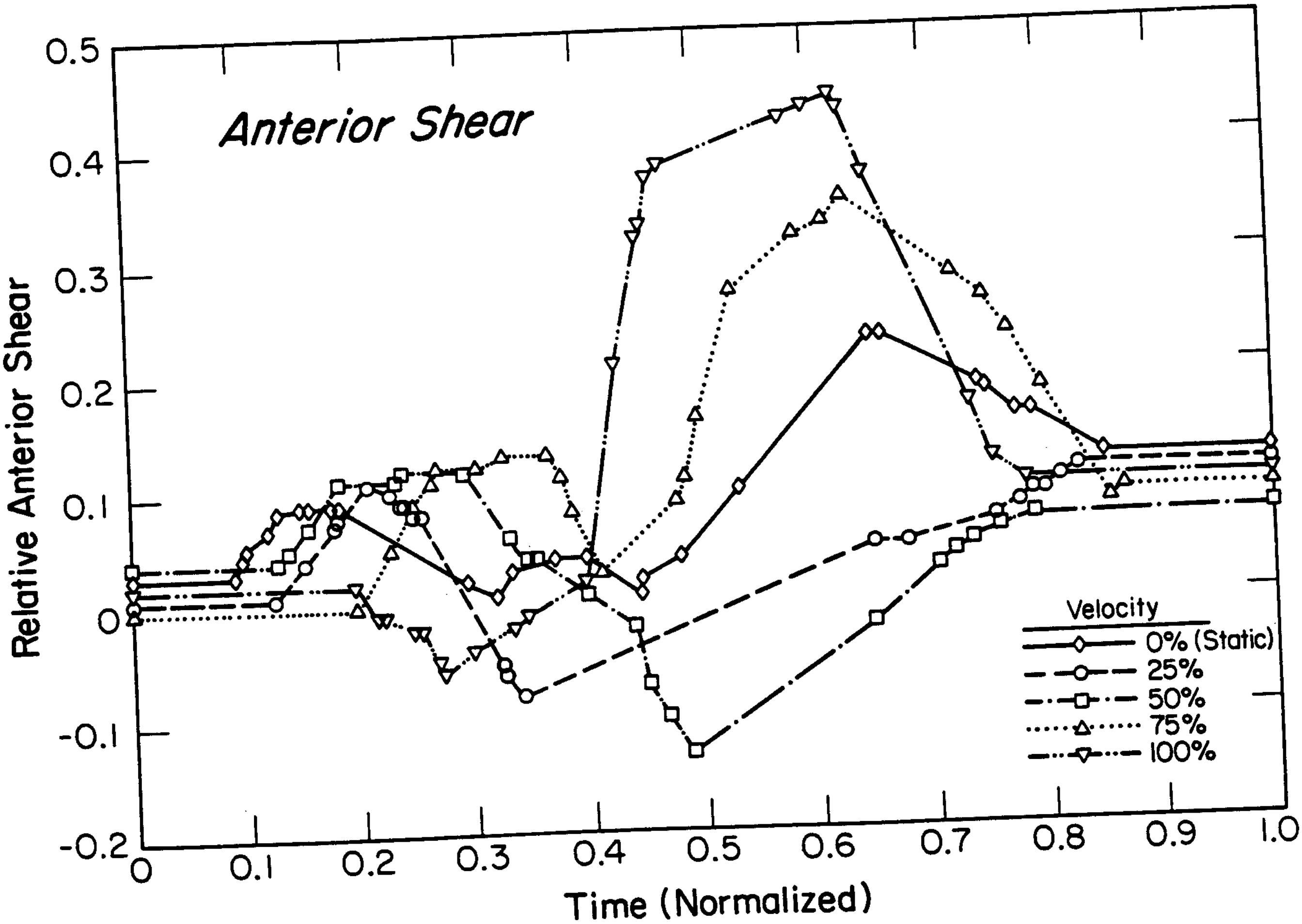

Fig 4. Plot of anterior shear index over simulated normalized time.

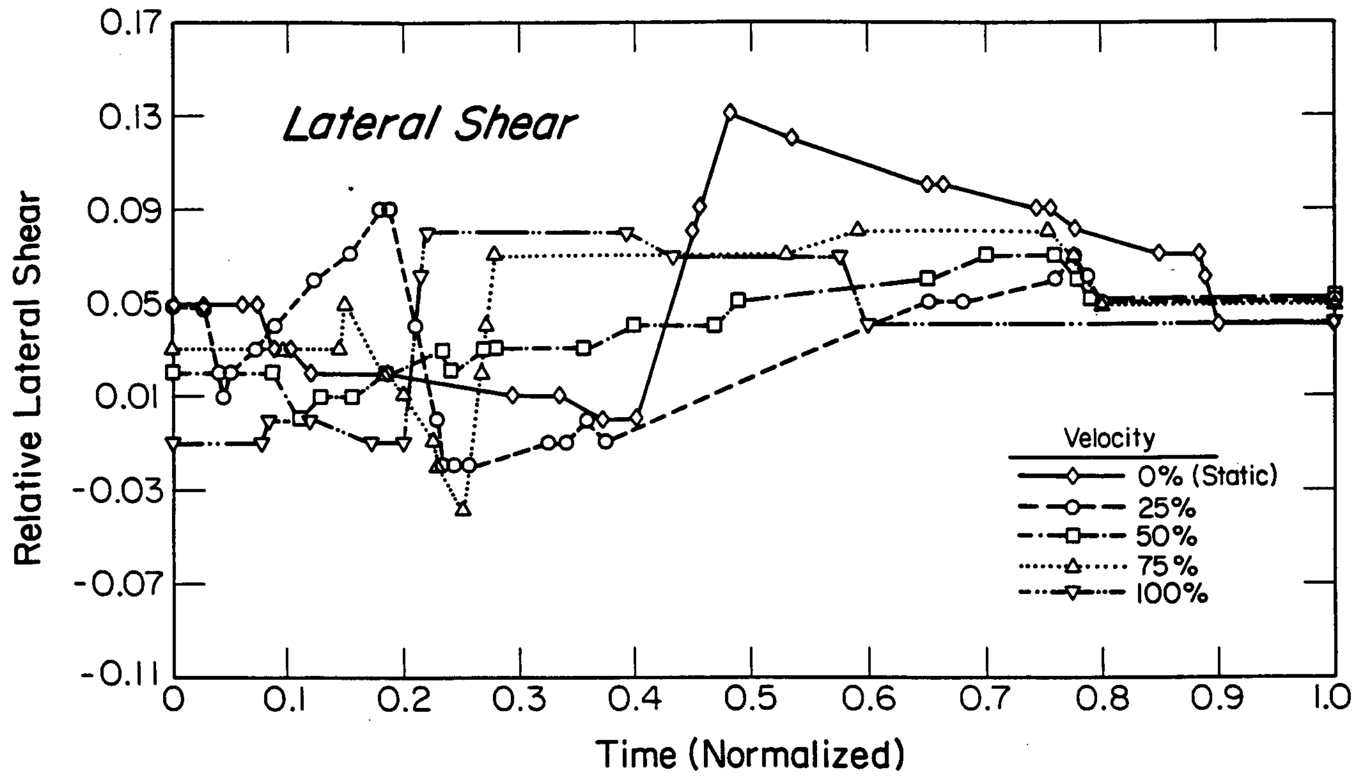

Fig 5. Plot of lateral shear index over simulated normalized time.

condition. (The values of the indices are assumed to change linearly over time between the times of lifting events.)

In Figure 3, we see a definite pattern in the index for compression. Compression increases after the start of the trunk motion, reaches a peak almost midway through the motion, corresponding roughly to peak activity for all muscles and peak IAP, and steadily declines as the muscles' activity and IAP decline. It appears that compression is lowest under the static condition, Condition 1. Compression tends to be highest for the low-velocity dynamic conditions (2 and 3). This observation is consistent with results reported recently by Marras et al.[5] Compression on the spine is slightly greater in the erect position than in the starting position, where the spine is bent forward at 67.5°.

The index for anterior shear also shows an interesting pattern. (Figure 4) As the trunk motion begins, there is a slight increase in anterior shear, followed by a decrease of modestly greater magnitude. Next, there is a more dramatic increase in anterior shear about midway through the motion. Finally, anterior shear reaches a level slightly higher than its original level. Anterior shear is greatest for dynamic Conditions 4 and 5 and least for dynamic Conditions 2 and 3. This may indicate that there is a tradeoff between anterior shear and compression between experimental conditions.

The trunk motion we are simulating is supposed to be a symmetric motion. Therefore, under all five experimental conditions, lateral shear should be zero throughout a simulated motion. We see in Figure 5, however, that this is not totally the case: lateral shear forces are observed. We observe this phenomenon because the ten subjects who provided our data did not perform the trunk motion in a truly symmetric manner. However, the calculated lateral shear is almost negligible, perhaps just "noise" (eg, muscle control error) in the system. The departure from symmetric motion is small enough as to have little effect on our simulation results.

Some of the summary statistics calculated by Simulift are displayed in Tables 2 and 3. The statistics presented in these tables are from the Simulift run for the 50% velocity condition. The average magnitudes of the profile components over the entire simulated time period, over the period of actual activity, and over the period that the profile element was engaged in activity attributable to trunk motion, are shown in Table 2. Table 3 displays summary statistics for the indices for compression, anterior shear, and lateral shear.

The entries in Table 2 indicate that the average proportion of maximum activity for each profile component that was observed

Table 2. Summary Statistics for Profile Components (50% Velocity Condition)

	Average proportion of maximum observed activity		
Profile component	*Over simulated time*	*During entire activity*	*Only while active*
LATL	0.36676	0.39932	0.46401
LATR	0.42618	0.46192	0.51413
ERSL	0.30673	0.34062	0.44744
ERSR	0.33594	0.36336	0.44720
EXOL	0.38704	0.41504	0.44883
EXOR	0.51663	0.53096	0.60467
INOL	0.48149	0.51050	0.56272
INOR	0.37769	0.39606	0.48298
RCAL	0.55767	0.57252	0.66186
RCAR	0.23428	0.25392	0.36149
IAP	0.41563	0.47060	0.48528

LATL = latissimus dorsi left; LATR = latissimus dorsi right; ERSL = erector spinae left; ERSR = erector spinae right; EXOL = external oblique left; EXOR = external oblique right; INOL = internal oblique left; INOR = internal oblique right; RCAL = rectus abdominus left; RCAR = rectus abdominus right; IAP = intra-abdominal pressure.

Table 3. Summary Statistics for Relative Indexes (50% Velocity Condition)

Statistic	Index for Compression	Index for Anterior shear	Index for Lateral shear*
Minimum	1.94443	−0.12971	0.00220
Maximum	4.56807	0.12114	0.07245
Averages†:			
Overall	2.82620	0.03145	0.04202
During activity	2.97878	0.02790	0.04427
LATL active	3.26700	0.01579	0.04140
LATR active	3.22432	0.01802	0.04260
ERSL active	3.41132	0.01138	0.04322
ERSR active	3.41007	0.01216	0.04012
EXOL active	3.16973	0.02106	0.04458
EXOR active	3.37181	−0.01694	0.05367
INOL active	3.25187	0.01657	0.04677
INOR active	3.52434	−0.01223	0.05094
RCAL active	3.94480	−0.02693	0.04500
RCAR active	3.68068	−0.01424	0.04738
IAP active	3.01447	0.02728	0.04569

LATL = latissimus dorsi left; LATR = latissimus dorsi right; ERSL = erector spinae left; ERSR = erector spinae right; EXOL = external oblique left; EXOR = external oblique right; INOL = internal oblique left; INOR = internal oblique right; RCAL = rectus abdominus left; RCAR = rectus abdominus right; IAP = intra-abdominal pressure.

*Values reported are for right-lateral shear.

†All averages are time-integrated averages.

during different periods of simulated time. Table 3 shows the contribution of the profile components to the indices for compression, anterior shear, and lateral shear, respectively.

In order to compare our impulse model with models that use EMG and IAP values that are averaged over time as most previous models do, we calculated (average) compression, anterior shear, and lateral shear indices using the average EMG and IAP values calculated by Simulift. Our findings indicate that the constant index values for compression, anterior shear, and lateral shear that are computed using time-averaged EMG and IAP readings are almost identical to the average values calculated by Simulift. The indices calculated on the basis of time-averaged EMG and IAP readings are shown in Table 4, as are the extreme values of the indices calculated by Simulift. By considering the time sequence of events in an impulse model like Simulift, we are able to capture the variability in the forces exerted on the spine during a lifting motion, not just the average values of the forces over time. This represents a distinct advantage of Simulift over those models that use time-averaged inputs. The entries in Table 4 and Figure 3 show that the change in the index for compression during trunk motion calculated with time-averaged inputs is roughly equal to 50% of the simulated change in the compression index; the discrepancy in the index for anterior shear is even greater. The variability and the extreme values of these forces are likely to have a substantial impact on the risk of low-back injury.

DISCUSSION

We have presented a deterministic simulation model, called Simulift, of human trunk motion. Although Simulift is a descriptive model, rather than a prescriptive model, it has several advantages over other human trunk models. First, trunk-muscle forces and IAP, which comprise a human trunk profile, are tracked over time; values based on averages taken over time are not used. Second, all 11 components of the profile can be active throughout the simulation. Third, exertions performed under both static and dynamic conditions can be simulated.

Simulift permits the time-dependent characterization, quantification, and analysis of the loading on the spine during trunk motion. Time-integrated statistics on the profile elements and indices for compression, anterior shear, and lateral shear, computed during a simulation, provide a convenient summary of the simulation results. Simulift is able to capture the instantaneous loading of the spine and the loading variability evident during dynamic exertions. This means that instantaneous spine loading may be quantified under dynamic exertions without invasive procedures like intradiscal pressure measurements.

We think that Simulift can be a valuable tool for analyzing human trunk motion during static, dynamic, and asymmetric exertions, given appropriate profile-component-pattern data. The evaluation of the trunk profile and the calculation of statistics need not be restricted to the times of motion events. They can be carried out at any point during the simulation, and as frequently as one may choose to in order to increase the accuracy of the summary statistics. It is also important to note that sensitivity analyses can be conducted by making several simulations with input-parameter sets that differ slightly.

Simulift is now being used to simulate absolute forces under certain anthropometric assumptions. The simulation results could be compared with cadaver spine tolerance data for ergonomic and design purposes. Magnetic resonance imaging (MRI) may be useful in better establishing the relationship between EMGs and actual forces; MRI can be used to determine the cross-sectional area of a muscle, and this area is related to the muscle's maximum force.

The present version of Simulift also can be embellished. It is clear that there is variability in the ways that different people exert themselves during trunk motion.[4] A stochastic simulation model that retains all of the basic features of our deterministic model is now being constructed. Motion-event times are generated randomly in accordance with the muscle-recruitment-pattern findings of Marras and Reilly and the observed mean values and variances of the motion-event times. The magnitudes of the re-

Table 4. Comparison of Simulift Indices and Indices Calculated Using Time-Averaged EMG and IAP Measurements

Experimental condition	Simulift minimum	Time-averaged data	Simulift maximum
	Compression index		
1	1.77350	2.72521	3.83743
2	1.89862	2.98953	4.43638
3	1.94443	2.82620	4.56807
4	1.92608	2.76590	4.30070
5	2.00418	2.80025	4.02264
	Anterior shear index		
1	0.00594	0.10023	0.23146
2	−0.09007	0.03278	0.11219
3	−0.12971	0.03146	0.12114
4	0.00424	0.12965	0.34852
5	−0.05550	0.14599	0.43674
	Lateral shear index		
1	−0.00009	0.05950	0.12986
2	−0.02089	0.03485	0.09139
3	0.00220	0.04202	0.07245
4	−0.03533	0.05360	0.08132
5	−0.10402	0.04016	0.08132

EMG = electromyography; IAP = intra-abdominal pressure.

sponse variables are also random. The means and variances of the observed values of the profile components are useful in determining parameters for assumed distributions of the profile components at motion-event times. The stochastic model will be particularly valuable in describing the effect of trunk-motion variability for one subject within an experimental condition.

The model extensions and embellishments suggested here are indicative of the potential of models like Simulift in the study of human trunk motion. Models like Simulift can provide us with insights that will lead us to a more thorough understanding of immediate and long-term effects of certain types of trunk motion.

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