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MECHANICS OF CPR PERFORMED WITH THE PATIENT ON A SOFT BED VERSUS A HARD SURFACE

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■ ABSTRACT

Objective: To study the effects of underlying bed softness versus stiffness on the effectiveness of chest compressions in CPR.

Methods: For a wide range of bed stiffness constants, mathematical models describing compression of the human chest supported by a hospital bed were created for an adult thorax experiencing either a sinusoidal compressive force or a sinusoidal sternal displacement.

Results: With 5 cm peak displacement, sternum-to-spine compression fell from 4.3 to 1.0 cm, and peak power fell from 59 to 23 Watts, as bed stiffness decreased from 50,000 to 5,000 N/m. Less than 35% of maximal chest compression occurred at a typical bed stiffness of 10,000 N/m. With 400 N peak force, sternum-to-spine compression decreased from 5.0 to 2.0 cm, and peak power increased from 82 to 226 Watts, as bed stiffness decreased from 50,000 to 5,000 N/m. However, greater than 85% of maximal chest compression was obtained at a typical bed stiffness of 10,000 N/m.

Conclusion: The deterioration of chest compression performed on soft beds is technique dependent. If necessary, CPR can be performed effectively on a softer surface using a constant peak force technique. However, a firm surface is most desirable.

Key Words: Bed, Cardiac Arrest, Cardiopulmonary Resuscitation, Chest Compression, Computers, Hospital, Mathematical Model, Mechanics

■ INTRODUCTION

Blood flow during CPR is linearly related to the depth of chest compression[1]. On a hard surface external chest compressions directly displace the sternum toward the spine, invoking either the cardiac pump or thoracic pump mechanisms to generate forward flow[2,3]. According to the 1992 guidelines for basic life support[4]: "If the victim is in bed, a board, preferably the full width of the bed, should be placed under the patient's back to avoid the diminished effectiveness of chest compression." For those patients who are large or who are connected to many monitoring and life support devices, the placement of a backboard can be difficult and time consuming. Sometimes the patient is moved to the floor, requiring interruption of CPR. These non-ideal situations raise the question of the risk versus benefit of performing CPR on a soft surface such as a hospital bed. Unable to fund a comprehensive study on the subject, we conducted a systematic mechanical analysis of the effects of substrate stiffness on chest compression in CPR.

■ METHODS

Mathematical model of patient bed mechanics: To capture the essence of CPR mechanics on a bed, we created a mathematical model of the chest-and-bed system using standard engineering principles[5], as shown in Figure 1. A relatively small mass, m_1 (~0.1 kg), representing the sternum is coupled to the remainder of the thoracic mass, m_2 (~30 kg), by elastic springs having a collective spring constant, k_1, and a damping element with damping coefficient, μ_1. The remainder of the thoracic mass rests upon an idealized mattress and a set of bedsprings, represented by parallel elastic elements, k_2, and mattress damping element μ_2.

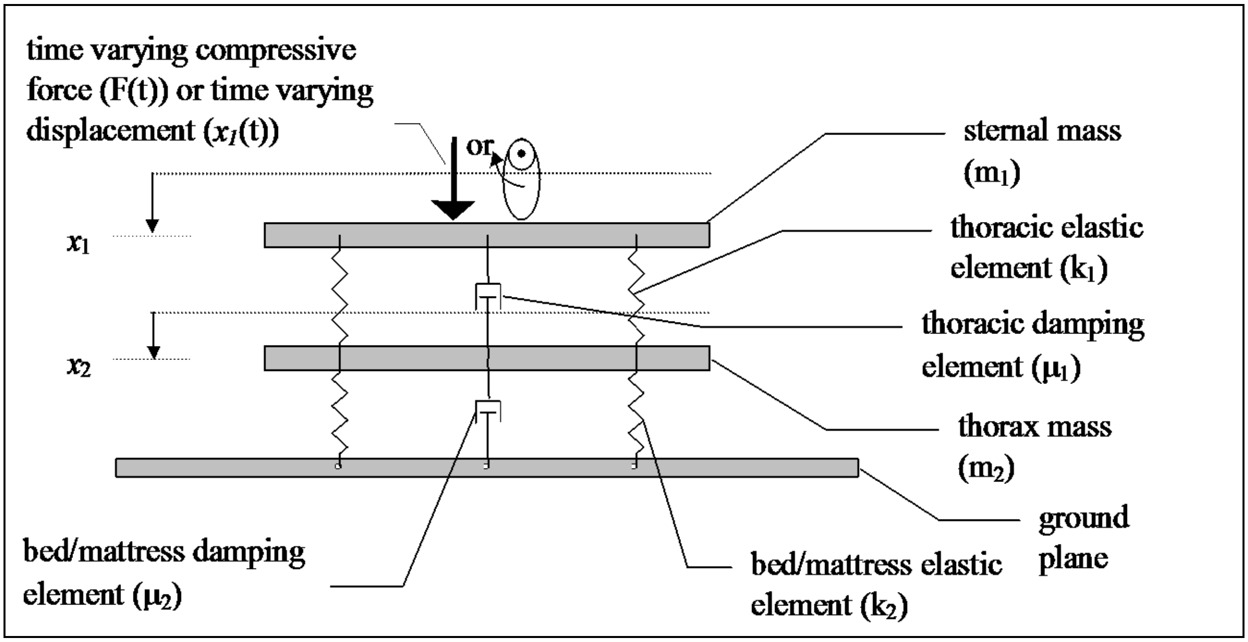

FIGURE 1. Illustration of the basic mechanical elements used to model the human chest - bed system.

Either a time-varying sinusoidal displacement or a time-varying sinusoidal force was used to drive the system. Compression of the "patient" on a bed during CPR causes the combined displacement of the sternum by a distance, x_1 , and displacement of the chest as a whole (i.e. the mattress surface) by a distance, x_2 , from their initial equilibrium positions. The net compression of the chest, which is related to the effectiveness of CPR, is the difference, x_1-x_2, between absolute sternal displacement and bed surface displacement. Thus, x_1-x_2 is the displacement of the sternum with respect to the spine.

In the CPR-related problem at hand, we are interested in the steady-state oscillations of this system when an approximately sinusoidal time varying compression is applied to the sternum. Our goal is to compute the effects of varying bed or substrate stiffness, k_2 , upon the net chest compression x_1-x_2. To solve this problem mathematically we made the following simplifying assumptions: (1)The sternal mass (~0.1 kg) is small with respect to the thoracic mass (~30 kg). (2) The springs and dashpots of the system are linear, once the bed has been pre-compressed by the weight of the patient. (3) The applied external force or displacement is sinusoidal (50% compression time, 50% relaxation time). (4) The oscillatory motions of the system have reached a steady-state, i.e. initial transients from start-up of CPR have dissipated. With these assumptions it is possible to solve analytically for x_1 and x_2 as functions of time for two modes of chest compression. In constant peak displacement compression, the sternum is displaced a fixed distance (e.g. 5 cm) regardless of any underlying movement of the chest and bed. In constant peak force compression, a sinusoidal time varying force is applied to the sternum, regardless of its downward motion, in an attempt to compensate for the softness of the underlying bed.

The equations of motion (force = mass x acceleration) governing the periodic fluctuations of unknowns x_1 and x_2 for either constant peak displacement or constant peak force chest compression on a bed or other soft surface are given by steady-state solutions of the following generalized second order differential equation:

$$A\ddot{x} + B\dot{x} + Cx = D\sin(\omega \cdot t) + E\cos(\omega \cdot t) + F \qquad (1)$$

where x is a function of time, t , ω is the angular frequency of compression in radians/sec, the symbol $\ddot{x}$ represents the second time derivative of x , the symbol $\dot{x}$ represents the first time derivative of x, and A, B, C, D, E, and F are constants in time, which depend on the model parameters and the compression mode. As can be confirmed by differentiation, the general steady-state solution of the above equation is the sum the sum of a sine term, a cosine term, and a constant. Table 1 shows the values of constants A, B, C, D, E, and F in expression (1) that correspond to specific solutions for the variables x_1 and x_2 for either constant peak displacement compression or constant peak force compression of the chest.

TABLE 1 Specific solutions of $A\ddot{x} + B\dot{x} + Cx = D\sin(\omega \cdot t) + E\cos(\omega \cdot t) + F$; where $x = a\sin(\omega \cdot t) + b\cos(\omega \cdot t) + h$, $a = \dfrac{D(C - A\omega^2) + BE\omega}{B^2\omega^2 + (C - A\omega^2)^2}$, $b = \dfrac{E(C - A\omega^2) - BD\omega}{B^2\omega^2 + (C - A\omega^2)^2}$, and $h = \dfrac{F}{C}$.

Coefficient	**x_1 (Constant Peak Force Model)**	**x_2 (Constant Peak Force Model)**	**x_2 (Constant Peak Displacement Model)**
A	0	m_2	m_2
B	μ_1	μ_2	$\mu_1+\mu_2$
C	k_1	k_2	k_1+k_2
D	$k_1a_2 - \mu_1b_2\omega$	0	$+\frac{1}{2}\mu_1\omega x_{1\max}$
E	$k_1b_2 - \mu_1a_2\omega - \frac{F_{\max}}{2}$	$-\frac{1}{2}F_{\max}$	$-\frac{1}{2}k_1x_{1\max}$
F	$\frac{F_{\max}}{2} + k_1h_2$	$+\frac{1}{2}F_{\max}$	$+\frac{1}{2}k_1x_{1\max}$

Solutions for constant peak displacement technique: In the specific simulation of chest compression with constant peak displacement, the sternum is pushed downward a fixed maximal distance, $x_{1\,max}$, regardless of the motion of the bed. The driving force x_1 as a function of time is defined as $x_1 = x_{1\max} \cdot \frac{1}{2}\left(1-\cos(\omega \cdot t)\right)$, with $x_1 = 0$ at time $t = 0$, and maximal value $x_{1\,max}$, whenever $\omega \cdot t = \pi$ radians. In response, the value of x_2 as a function of time, t, is given by a particular steady-state solution of equation (1), indicated in the rightmost column of Table 1. Knowing both x_1 and x_2, it is possible to compute the sternal compression with respect to the spine, $x_1 - x_2$, during constant peak displacement CPR for multiple cases in which mattress firmness, k_2, varies.

Solutions for constant peak force technique: In the specific simulation of compression with constant peak force, the sternum is pushed downward with a defined sinusoidal force, regardless of the "give" of the underlying bed. That is, the same force is applied to the sternum even if the chest as a whole recedes with compression of the bed. Here applied force as a function of time, $F(t) = F_{\max} \cdot \frac{1}{2} \cdot (1-\cos(\omega \cdot t))$, such that $F(0) = 0$, and $F(t) = F_{max}$ periodically whenever $\omega \cdot t = \pi$ radians. The value of x_2 is given by the solution of the second order differential equation (1), when constants A through F are determined as shown in the middle column of Table 1. Having found solutions for x_2 as a function of time, the solution for x_1 with constant peak force compression can be computed from (1) when constants A through F are defined as shown in the leftmost column of Table 1. In this way we could compute true sternal compression with respect to the spine, $x_1 - x_2$, for either constant peak displacement or constant peak force compression, given reasonable experimental values for the spring constants, k, and damping constants, μ, of typical adult chests and hospital beds.

Spring and damping constants of the chest: Gruben et al.[6] have previously characterized the elastic and damping properties of the human chest. We chose their values for typical chest stiffness and damping coefficients for the model of Figure 1. With the elastic constant of the chest equal to 7500 N/m and the damping constant of the chest equal to 275 N/m/s, the model reproduced the measured force displacement data for the human chest, including hysteresis.

Spring and damping constants of the bed and mattress: To estimate experimentally the values of the spring constant, k, and damping coefficient, μ, for a hospital bed under test, one can measure the resonant angular frequency, ω, when a series of test masses, m, is applied to the bed and the system is set into damped oscillatory motion. For test mass, m, placed on a bed with stiffness k and damping μ

$$\omega = \frac{1}{2 \cdot m} \cdot \sqrt{4 \cdot k \cdot m - \mu^2} \quad , \quad \text{where } \mu^2 < 4 \cdot k \cdot m \text{ (under-damped case)}^5. \qquad (2)$$

From expression (2) the spontaneous resonant angular frequency, ω, of the system when the mass is "let go," is dependent on constants k and μ, such that $m^2 \cdot \omega^2 = k \cdot m - \frac{\mu^2}{4}$. In reality, the mattress gets stiffer with increasing weight of compression. If one assumes a simple linear increase in spring constant, k , with slope, b , as a function of the weight, $m \cdot g$, of the preload (i.e. $k(m) \cong k_0 + m \cdot g \cdot b$), then (2) becomes a quadratic equation in *m*,

$$m^2 \cdot \omega^2 = -\frac{\mu^2}{4} + k_0 \cdot m + g \cdot b \cdot m^2 . \qquad (3)$$

Experimentally, we placed several known masses on a "chest sized" board (dimensions: 30 x 44 cm) on the bed to be tested. The board and mass were then displaced downward. At the moment of their release, a stopwatch was started and the number of cycles of upward movement was counted until motion was no longer detectable (< 1 mm of displacement). The total number of cycles was divided by the total time for all of the cycles to obtain the frequency of the harmonic motion (cycles per second). This frequency was then converted to an angular frequency for each mass tested ($\omega = 2 \cdot \pi \cdot \text{frequency of oscillation}$) and the values of $m^2\omega^2$ were plotted as a function of m, from which $k(m)$ and μ could be estimated from regression analysis (Figure 2). The test masses were a 22 kg sack of sand and people of varying body weights (31, 48, 66, 80 and 88 kg), determined on a Detecto Model CN20 digital scale. These people sat down on the target 30 x 44 cm board, placed as close as possible to the center of the tested hospital bed, to initiate damped oscillations of the bed-mass system. The oscillations were counted and timed with a stopwatch as long as they were clearly perceptible, while the "test mass" endeavored to remain still and balanced on the board. A second order polynomial curve fit was applied to the data for $m^2 \cdot \omega^2$ as a function of m in order to evaluate $k(m)$ and μ , as shown in Figure 2. In this manner values of k and μ were estimated experimentally for four in-service hospital beds with and without CPR-backboards, using only a stopwatch, a wooden board, human volunteers of different size, and a bag of sand.

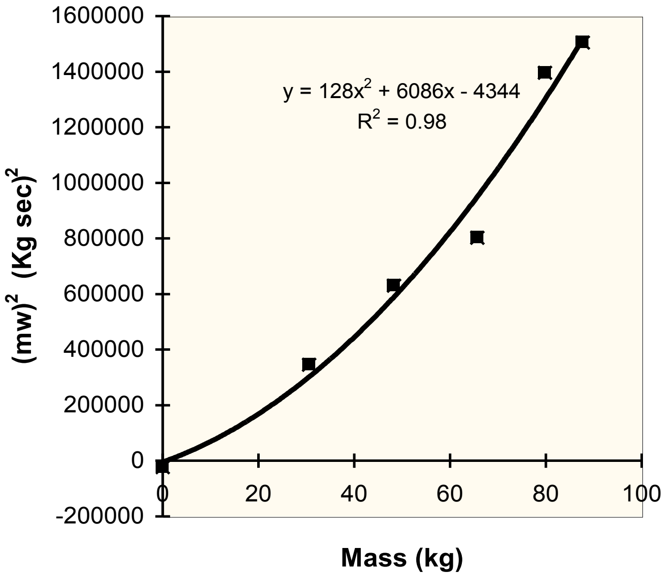

FIGURE 2. Example of $m^2\omega^2$ versus m plot for the Hill-Rom 720 bed and the May & Co. Mattress. Solid data points represent measured values. Smooth curve represents least squares regression used to estimate bed stiffness and damping constants. The experimental value of $\mathrm{m}^2 \cdot \omega^2$ **at *m* = 0 was taken as** $-\frac{\mu^2}{4}$ **for the value of the damping constant,** μ **, which reproduced the observed duration of detectable oscillations in the tested bed when inserted into the expression** $x(t) = x(0)\, e^{-\mu t/(2m)} \sin(\omega\, t)$.

Computed Solutions for chest displacement: Using values for chest and bed stiffness constants and damping factors, obtained as just described, and expressions for $x_1(t)$ and $x_2(t)$ for both constant peak displacement compression and constant peak force compression of the chest, we conducted a spreadsheet analysis using *Microsoft Excel 6.0* to explore the effects of system parameters upon sternum-to-spine compression. The accuracy of mathematical expressions in the spreadsheet program was validated using 16 simple test cases for which the results were known or for which there exists a simple analytical solution. Models behaved as predicted for all test cases.

■ RESULTS

The damping coefficients and spring constants obtained from a variety of bed and mattress combinations are summarized in Table 2. The last column of this table gives a sample spring constant for the bed/mattress combination with the application of a 30 kg mass as was used in our modeling. Backboards greatly increased the overall stiffness of the bed/mattress combination and also caused an increase in damping. We chose to use the Hill-Rom 720 bed with the May & Co. mattress as the standard bed/mattress combination for subsequent modeling.

TABLE 2 Experimentally determined bed and mattress constants. Values are averages for 2 or 3 trials.

Bed Type	**Mattress Type**	**k_0 (N/m)**	**b (N/m/kg)**	**μ_2(N/m/s)**	**k_2 (N/m) with a 30 kg mass**
Hill-Rom 720	May & Co.	6086	13.2	132	9970
Hill-Rom 720	May & Co. + backboard	8884	13.5	335	12860
Hill-Rom 720	Gel Foam	9256	17.2	503	14320
Hill-Rom 450	May & Co.	4271	16.2	198	9040
Hill-Rom 450	May & Co. + backboard	12240	13.7	519	16270
Hill-Rom 450	Gel Foam	8475	20.5	545	14510

Figure 3 is a sample plot of the computed fluctuations in sternal displacement with respect to the spine (x_1-x_2) during CPR at 90 compressions/minute on a mattress of average stiffness (12,000 N/m) for the constant peak force and constant peak displacement models. Peak compression for the constant peak displacement model is approximately 3 cm under these conditions, while that for the constant peak force model is approximately 5 cm.

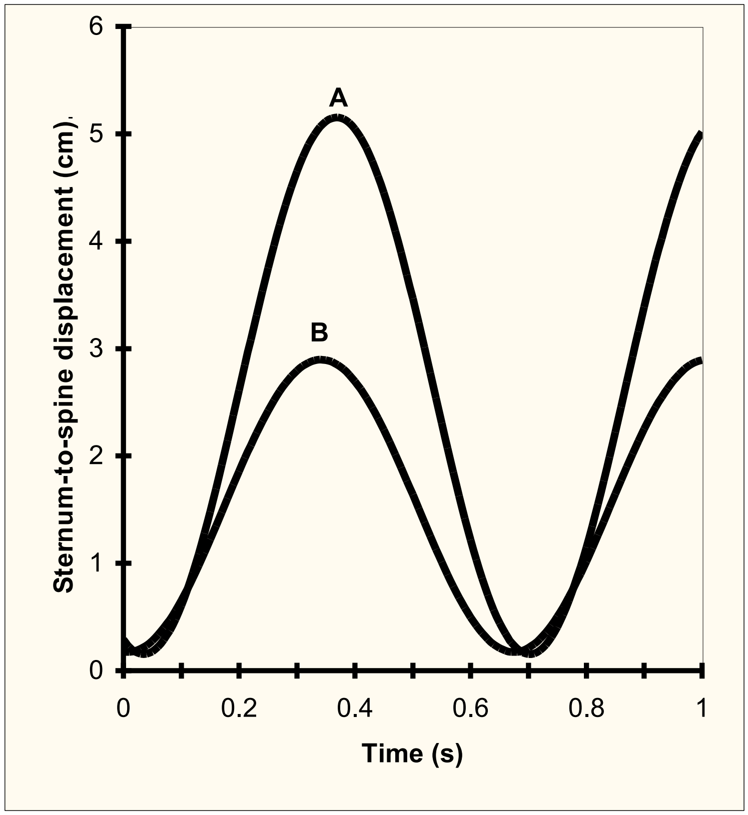

FIGURE 3. Time variance of sternal displacement with respect to the spine for constant peak force model (A) and constant peak displacement model (B) for CPR performed at 90 compressions/minute, a mattress stiffness of 12,000 N/m, and a mattress damping coefficient of 165 N/m/s.

Figure 4 summarizes computed chest displacement and peak power requirements (force times velocity) for each model as a function of bed stiffness. Within the dashed "normal" range in Figure 4, representing measured hospital mattress stiffness constants, the constant peak force technique is capable of generating a much higher degree of chest compression than the constant peak displacement technique. However, power requirements for the constant peak force technique are significantly greater than those for the constant peak displacement technique.

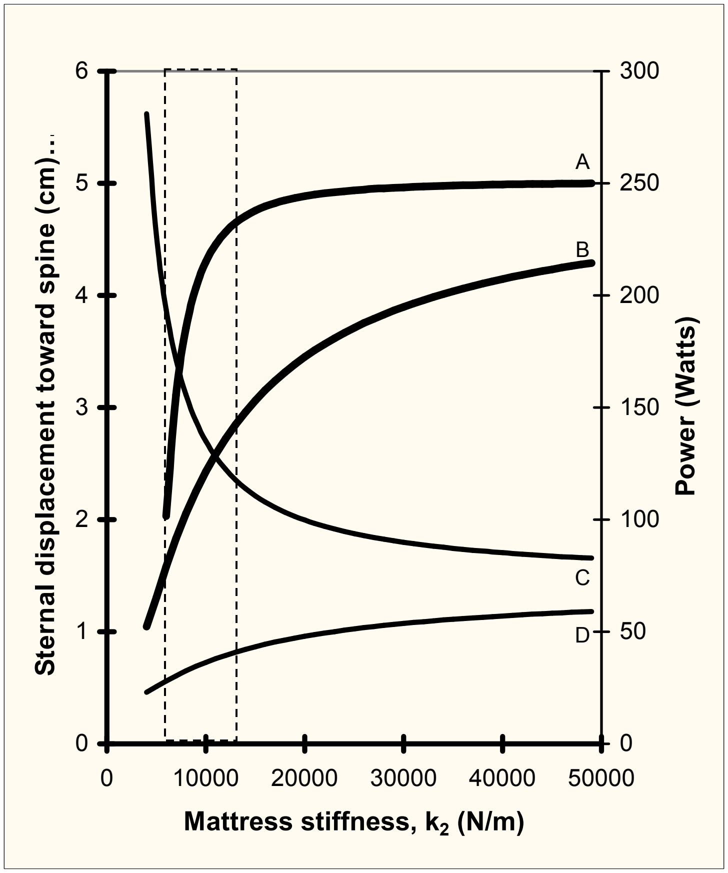

FIGURE 4. Chest displacement toward the spine and power requirements at varying mattress stiffness for CPR performed at 90 compressions/minute with either a constant peak force of 400 N or a constant peak displacement of 5 cm. A: sternum-to-spine compression, constant peak force model; B: sternum-to-spine compression, constant peak displacement model; C: power, constant peak force model; D: power, constant peak displacement model. Dashed vertical lines indicate the approximate normal range of hospital mattress stiffnesses.

■ DISCUSSION

Current CPR guidelines instruct the rescuer to depress the sternum 4-5 cm with each compression. Such wording is quite appropriate and unambiguous, if the patient is lying on a hard surface. On a soft surface, however, there are at least two possible interpretations. With constant peak displacement technique the rescuer's hands always travel a fixed distance and may tend to compress the bed rather than the chest. On soft surfaces the overall reduction in actual chest compression using this technique is great and the consequences to the patient are grave.

Any method increasing the stiffness of an underlying mattress, such as use of a backboard, will increase the degree of chest compression and, hence, the efficacy of CPR. A feasible alterative is constant peak force technique, with which the rescuer compresses the sternum using the same maximum force regardless of any patient motion. This mode is similar to that applied by the Thumper® mechanical resuscitator, and also by smaller adult rescuers who focus on using upper body weight to apply chest compressions. The constant peak force technique is capable of maintaining a significant degree of chest compression on all but the softest surfaces, at the expense of greatly increased work by the rescuer.

■ LIMITATIONS AND FUTURE QUESTIONS

The present study has been done with a mathematical model incorporating linear differential equations to approximate the non-linear performance of a chest and bed compressed during CPR. Further research should characterize specialized sleep surfaces and Gurneys as well as non-linear breakdown of the chest during prolonged CPR.

■ CONCLUSIONS

The effectiveness of chest compression during standard CPR may be seriously degraded on soft supporting surfaces such as hospital beds. The degradation may be reduced by the use of backboards and, if necessary, a constant peak force compression technique.

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