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# INTEGRATED MEDICAL ANALYSIS SYSTEM

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## ABSTRACT

This paper describes the Integrated Medical Analysis System. This evolving system consists of an integrated suite of models and tools providing quantitative and dynamic analysis from physiological function models, clinical care patient input, medical device data, and Northrop Grumman medical products. The System is being developed for requirements definition, testing, validation, control theory, and real-time diagnostic insights.

Unique system integration of components is achieved. The current prototype emphasizes cardiovascular and pulmonary physiological functions and integration of patient device data. An overview of the project and preliminary findings are introduced.

## 1 INTRODUCTION

The Integrated Medical Analysis System (IMAS) forms a complex analytical tool with emphasis on integration and interaction at multiple levels between components. This unique level of integration and interaction facilitates quantitative analysis for multiple purposes and varying levels of fidelity.

There are four facilitators of this system. These facilitators include a distributed interactive computing architecture, application of fluid and structural engineering principles to the models, real-time graphical data analysis, and application of strong system integration principles.

This paper will begin with a discussion of motivation for this work and projected uses and objectives of IMAS. Integrated system components are discussed with emphasis in this paper on the cardiovascular physiological model and on the distributed computing architecture. Distinctions of the cardiovascular model are presented. Finally, we will present conclusions and further research.

## 2 MOTIVATION

Traditionally, human models and simulations are performed on small scale, isolated problems, usually consisting of detached mathematical models or measurements studies. These systems are not capable of portraying the interactive effects of such systems and certainly are not capable of integrating multiple external entities such as device data, patient data, etc. The human body in and of itself is an amazing feat of system integration. External monitors, treatments, and medical conditions interact at yet another level. Hence, a highly integrated, interactive simulation system is required for effective quantitative analysis.

Until recently, other obstacles hindering such system development has included the underlying computing architecture. Distributed processing has made such a supportive infrastructure plausible. Graphical data analysis not only promises more immediately readable analysis, but also provides the opportunity to fine-tune the system models. Recent advancements in graphical packages have also extended graphical data analysis capabilities.

System objectives and potential applications revolve around two focuses:

- Requirements definition and system evaluation
- Rapid patient assessment insights.

With the representation and interactive integration of human patient models, process, treatments, and device interaction, we are then able to test and evaluate device and/or product performance. These scenarios also allows us to define device or product requirements necessary for our production methods.

For rapid patient assessment, basic patient parameters that are currently available non-invasively from physiological monitors provide limited parameters and an extremely limited view of the human status. The distributed interactive system allows us to *individualize* the simulation to particular patient characteristics and then perturb the system with parameter changes,

producing analytical results. We term this process *patient-in-the-loop* or *device-in-the-loop* simulation. With the integration of the individual patient characteristics, of those limited interactively monitored values into a whole human modeling infrastructure; we become able to present the *bigger picture* of what is occurring in that particular patient with a reasonable degree of accuracy. We are also able to reflect clinical history data through our database and brokering schemes. Furthermore, once future decision support is established with risk factors and trend analysis data, Integrated Medical Analysis System will become increasingly capable of providing additional patient assessment insights to medical personnel. We have likened this auxiliary assessment information for medical personnel to flight information provided to the cockpit pilot.

## 3 INTEGRATED SYSTEM COMPONENTS

Key aspects of the Integrated Medical Analysis System include the physiological models, device data brokering, data infusion, distributed computing architecture and graphical data analysis. The technology transfer of experienced fluid and structural mechanics principles have contributed to development and advancement of physiological models. Under the current version, the cardiovascular model is the cornerstone of IMAS and so will be discussed in the next section.

### 3.1 Cardiovascular model

The cardiovascular model in IMAS is characterized as a fluid dynamics problem coupled with an elastic, pressurized tube (a vessel). The theoretical model is based on several long-standing laws from engineering mechanics. These laws define a system of equations that can be programmed and solved in a time domain. A half sine wave was used as the forcing function to define pressure and flow rates leaving the left ventricle. A first order, backward finite difference algorithm was used to numerically solve the coupled derivative equations Fixed time steps of 0.001 seconds were used.

#### 3.1.1 Governing Engineering Mechanics Laws

- First Law of Thermodynamics:

Conservation of Energy
Bernoulli Equation

- Conservation of Mass:

Definition of a Continuum.

- Newton's Second Law of Motion:

Momentum Principles.
Navier-Stokes Equation

#### 3.1.2 Blood And Blood Flow Modeled as

- Blood is a viscous, incompressible Newtonian fluid.
- Vessels (arteries and veins) are pressurized elastic tubes with thick or thin walls. The walls are tapered or parallel.
- Vessel walls are modeled as linear elastic, Hookian material that follows the Kirchhoff-Love (Cook,Young) plate theory of thin membranes.
- Blood flow state is one dimensional, irrotational, axisymmetric, fully developed laminar flow.
- Velocity profile is parabolic.
- The high order terms and convective terms in the Navier-Stokes equations can be neglected.
- Nonlinear behavior is represented by piecewise linear discretization.

#### 3.1.3 ViscoElastic Coupled Model

Figures 1, 2 and 3 present concept models of which our cardiovascular model is designed. Figure 1 depicts the overall control volume model for a segment. Figure 2 depicts the electrical circuit model of an arterial vessel for which our equations revolve around. Figure 3 depicts the visco-elastic mechanics model for an arterial vessel in IMAS.

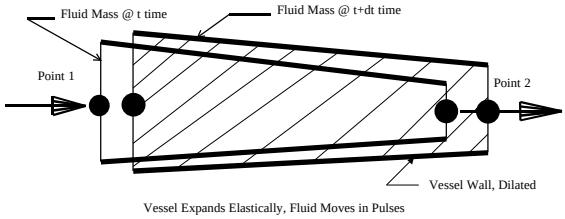


Figure 1. Control Volume Model

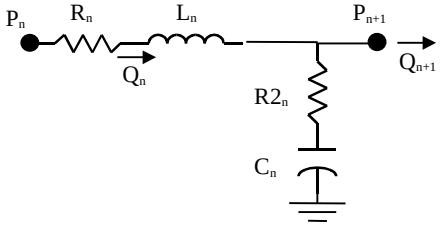


Figure 2. Electrical Circuit Model of Arterial Vessel (Snyder,Rideout)

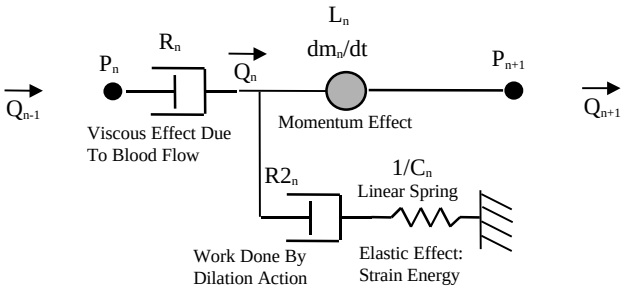


Figure 3. Visco-Elastic Mechanics Model of Arterial Vessel (O'Neil)

Figure 4 provides multiple velocity waveforms as blood flow decends from the aorta into major arterial segments over time.

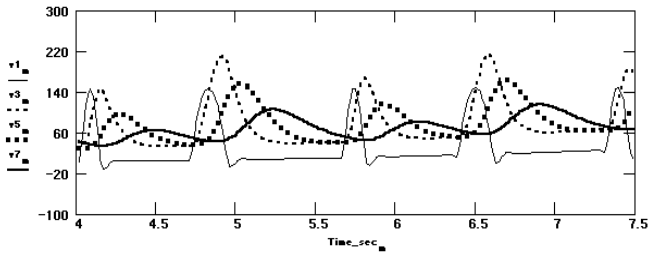


Figure 4. Velocity Waves of Flow through Descending Arterial Segments (Cm/Sec).

### 3.1.4 Governing Equations

*I. Generalized Navier-Stokes Equation:*

$$\rho\frac{D\vec{V}}{Dt} = -\nabla P + \vec{B} + \mu\nabla^2\vec{V} \qquad (1)$$

Where

| | | |
|---|---|---|
| P | = | pressure |
| **V** | = | Velocity vector |
| **B** | = | Body force vector |
| D/Dt | = | Material derivative |

*Expanded Navier-Stokes Equation* (Roberson,Crowl)*:*

For irrotational flow:

$$\frac{D}{Dt} = \frac{\partial}{\partial t} + V_r\frac{\partial}{\partial r} + V_z\frac{\partial}{\partial z}$$

Material Derivative

$$\nabla\bullet V = \frac{1}{r}\frac{\partial}{\partial r}\left(rV_r\right) + \frac{\partial V_z}{\partial z}$$ Divergence

$$\nabla^2 = \frac{\partial^2}{\partial r^2} + \frac{1}{r}\frac{\partial}{\partial r} + \frac{\partial^2}{\partial z^2}$$ Laplacian

*Longitudinal Component:*

$$\rho\frac{DV_z}{Dt} = B_z - \frac{\partial P}{\partial z} + \mu\nabla^2 V_z \qquad (1b)$$

Assuming there are no body forces and that higher order terms can be dropped (linearize). Then:

$$\rho\left[\frac{\partial V_z}{\partial t} + V_r\frac{\partial V_z}{\partial r} + V_z\frac{\partial V_z}{\partial z}\right] = -\frac{\partial P}{\partial z} + \mu\frac{1}{r}\frac{\partial}{\partial r}\left\{r\frac{\partial V_z}{\partial r}\right\} \qquad (1c)$$

This equation balances the momentum forces (due to motion of the fluid) with the "surface traction" forces (external forces acting at the surface of the control volume.) (Roberson,Crowl)

From the definition of parabolic velocity profile, the definition of shear, and the definition for Q and dropping radial components (variations in radial direction):

$$-\frac{\partial P}{\partial z} = \hat{L}\frac{\partial Q}{\partial t} + \hat{R}Q \qquad (2)$$

Ye, Moore and Jaron showed that this equation can be converted for a parabolic velocity profile such that the coefficients have the following definitions:

$$\hat{L} = \frac{\rho}{\pi r_o^2}$$ Inductance, Momentum (2a)

$$\hat{R} = \frac{8m}{pr_o^4}$$ Flow Resistance (2b)

Where $\hat{L}$ is the equivalent fluid inductance, and $\hat{R}$ is the equivalent fluid resistance. The pressure term on the left defines a pressure gradiant and is a negative number indicating that pressure is lost as fluid flows in a vessel. ¶*z* is the length of the vessel. This equation balances the pressure lost (as the fluid flows in the vessel) with the change of momentum plus the shear stress at the vessel wall that resists the motion of fluid.

Eq. 2 can be rearranged to form:

$$¶Q = -\frac{1}{\hat{L}}\frac{¶P}{¶z}¶t - \frac{1}{\hat{L}}\hat{R}Q¶t \qquad (2c)$$

The code for this equation is:

```
artery.dFa[n_seg] = (-( artery.Pa[n_seg] -
    P )/val.artery_length [n_seg] - R *
  ( artery.Fa[n_seg] + F ) / 2.0 ) * dt / L;
artery.Fa[n_seg] +=  artery.dFa[n_seg];
```

*II. Bernoulli Energy Equation:*

From classic fluid mechanics we obtain the following energy equation for a pump system in a closed loop pipe system with varying diameters (the pump is after node 1

and before node 2, flow is fully developed, laminar and fills the pipes):

$$\frac{P_1}{g}+\frac{V_1^2}{2g}+z_1+H_{pump}= =\frac{P_2}{g}+\frac{V_2^2}{2g}+z_2+\Sigma K_L\frac{V^2}{2g}+\Sigma\frac{fL}{D}\frac{V^2}{2g} \quad (3)$$

Where:

$P_1$ = Pressure at point 1 (before reaching the pump)
$V_1$ = Velocity of fluid at point 1.
$Z_1$ = Spatial position (in Z direction) of point 1
$H_{pump}$ = Energy input into the system by the pump
$P_2$ = Pressure at point 2 (after the pump)
$V_2$ = Velocity of fluid at point 2.
$Z_2$ = Spatial position (in Z direction) of point 2
$H_{pump}$ = Energy input into the system by the pump
$K_L$ = Kinematic factors that cause energy loses
fL = Internal friction factors that cause energy loses as the fluid moves in the pipe.

$$f=\frac{64}{Re}$$

$$Re=\frac{\bar{V}D}{\nu} \quad \text{(Reynolds Number)}$$

In (Roberson,Crowl) the generalized Bernoulli energy equation is presented. A simplified Bernoulli equation was installed in code as:

```
P1 = P0 / density_blood + ( V0*V0 - V1*V1 ) /
        2.0 / g_cm_sec2;
```

*III. Elastic Dilation of Vessels:*

From the mechanics of elastic cylinders (2), we have the following definitions for thin and thick walled vessels:

*THIN WALL VESSELS (h/r < 10%)*

$$\delta Vol=\frac{3}{2}\frac{\pi r^3}{Eh}L\delta P$$

This equation defines the change in volume of a vessel for a given pressure change (vessel walls are parallel and thin, fluid is stationary) . The pressure change **P** is the effective pressure acting which is the outside pressure minus the inside pressure. **h** is the thickness of the vessel wall. **r** is the radius to the inner wall surface.

This can be rearranged to:

$$dQdt = C_n\Delta z dP \quad (4)$$

Where

$$C_n=\frac{3\pi r_n^3}{2E_n h_n} \quad \text{(Elastic Coefficient of wall)}$$

Let $R_{2n}$ = correction for energy loss in the wall:

$$R_{2n}=\frac{0.002}{C_n\Delta z} \quad (4a)$$

Then, the elastic equation can be adjusted for energy loss:

$$dP=\frac{1}{C_n\Delta z}\left(\int dQdt + R_{2n}dQdt\right) \quad (4b)$$

This was coded in C as:

```
dPa = artery.dFa[n_seg] * ( dt + 0.002 ) / elastic()
                / val.artery_length[n_seg];
artery.Pa[n_seg] += dPa;
```

*THICK WALL VESSELS (h/r >= 10%)*

$$\delta Vol=\frac{2\pi r^2}{E}\left(\frac{a^2+b^2}{b^2-a^2}+\nu\right)L\delta P \quad (4c)$$

Where:

a = inside radius of vessel
b = outer radius of vessel (to outer layer of wall)
ν = Poisson's ratio (0.5)

Cook has shown the thick-wall solution converges to the thin-wall solution as h/r approaches 10%. The change in flow Q is related to the change in volume and thereby, to the change in radius. Using calculus of variation (Hildebrand), it can be shown that

$$\delta Vol = 2\pi\Delta z h\delta r \quad (5)$$

The code for this equation is:

```
dr = artery.dFa[n_seg] * dt / 2.0 / pi /
  val.artery_length[n_seg] / rad_now;
```

*IV. Poiseuille-Hagen Law:*

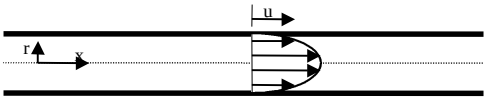


Figure 5. Parabolic Velocity Profile

For fully developed, laminar flow, for a rigid wall vessel, with parallel walls, and for a velocity profile that is parabolic, the internal friction losses will be given by:

$$C_{R_j}=\frac{8m}{pr_j^4}=\frac{128m}{pD_j^4}=\hat{R} \quad (5b)$$

$$Q=-\frac{\pi r_{max}^4}{8\mu}\frac{dP}{dz} \quad (6)$$

This equation defines a "resistance" value for vessel segment "j". Blood in-vivo, however, does not always meet the conditions stated. So this law needs to be used with discretion.

*V. Continuity Equation:*

Streeter presents the following continuity equation applied to a control volume:

$$\frac{1}{A}\frac{dA}{dt} = -\frac{\partial V}{\partial z}$$

Which relates the dilation of the vessel to the axial velocity of the fluid. In cylindrical coordinates *t*he equivalent equation is (Boresi, Chong):

$$\frac{1}{r}\frac{\partial}{\partial r}\left(rV_r\right) + \frac{\partial}{\partial z}V_z = 0 \qquad (7)$$

Figure 6 provides a 3D graphical analysis of the velocity waves as the blood flow progresses through the aorta and decending arterial segments over time. This plot is the 3D version analagous to the 2D comparison presented in Figure 4.

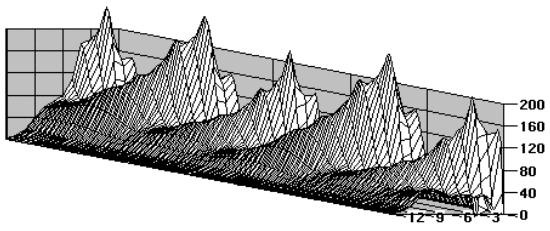


Figure 6. Velocity (Cm/Sec) in Descending Arterial Tree.

## 4 DISTRIBUTED COMPUTING ARCHITECTURE

Underlying IMAS is a distributed interactive computing architecture supporting distribution of components across multiple platforms, distribution of processes, communication and necessary synchronization between processes, user interface, and real-time graphical analysis.

Related to the distributed computing infrastructure is the data brokering employed for data access and data navigation over a heterogeneous, distributed network. This allows a more flexible and extensible system through a distributed organization of data and distributed access to that data.

The Northrop Grumman medical data archive contains data available to IMAS. The data pertains to the events and signals generated by the devices themselves as well as monitored patient status. Largely, the schema is inclusive of those devices that are components of the Northrop Grumman Life Support for Trauma and Transport (LSTAT$^{TM}$). Such devices (i.e. ventilator, defibrillator, drug infusion system, blood chemical analyzer, and physiological monitors) are essential for supporting and monitoring a critical care patient. Detailed device data is used for analysis in IMAS.

Access of this data is available through conventional software as well as object request brokers for system navigation over the heterogeneous, distributed simulation system. The object brokers are designed to shield the requester from the details of access and data location.

An object oriented database management system (OODBMS) serves as the IMAS database, residing on a Sun server having persistent objects. Distributing the clients and servers allows broker remote clients to perform function calls to the OODBMS persistent objects. In addition, the remote client level objects (via an adapter) become representation copies of the OODBMS objects. This powerful feature enhances the distributed functionality by providing the properties of persistency in the remote clients and provides the OODBMS objects with the distributed properties of an object request broker.

## 5 CONCLUSION

In this paper we have introduced the prototype Integrated Medical Analysis System. We have demonstrated the concepts and simulation capabilities within the context of this integrated suite of tools. Governing equations and principles applied in the cardiovascular physiological model has been presented. In addition to the general objectives of the system, implementation has been discussed of the data brokering techniques employed to integrate physiological device data and patient status to the simulation system. Future work involves the development of additional models, 3D anatomical graphics, supportive knowledgebases and decision support. The integration of multiple physiological models, dynamic patient state changes, data brokering over a distributed computing architecture and real-time graphical data analysis results in a powerful simulation and analysis system for health care environments.

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