Visualization and Quantification of Blood Flow in the Human Aorta. From in vivo 4D Phase Contrast MRI to Subject-Specific Computational Hemodynamics

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Insight into the Physiological Relevance of Helical Blood Flow in the Human Aorta

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Blood flow in the aorta is highly complex. In the past, massive observations demonstrated that helical flows predominate in areas from the ascending aorta to the aortic arch (Segadal & Matre, 1987; Kilner et al., 1993; Chandran 1993).

However, there is a relative paucity of in vivo quantitative data regarding helical blood flow dynamics in the human aorta. That helical flows predominate in areas from the ascending aorta to the aortic arch (Segadal & Matre, 1987; Kilner et al., 1993; Chandran 1993).

That this form of blood flow is a basic pattern for almost all the subjects no matter age and gender.

(Bogren & Buonocore, 1999; Houston et al., 2003)
It has been proposed that energetic constraint is but one consequence of the process of physiological evolution of helical blood flow in aorta, and that others remain to be discovered.

However, there is a relative paucity of quantitative data regarding helical blood flow dynamics in the human aorta.

**Qualitative Observations**

**NOT QUANTITATIVE**
Rationale, Aim, How

Rationale
Study of mechanistic relationship between physiological complexity and energy of aortic flow

Aim
Identify common features in physiological aortic bulk flow topology

How
In vivo aortic helical flow quantification in 5 healthy humans by applying 4D PC MRI
By using a Lagrangian representation of the aortic flow, we apply an index for helical flow quantification
4D PC MRI Data Acquisition

- Five healthy volunteers (men; age 23-42 years)
- HR range 43-78 bpm
- Philips Achieva 1.5 T scanner (Philips Healthcare)
- TR=5.4 ms, TE=3 ms, flip angle=15°, velocity encoding = 150 cm/s
- Navigator-echo to further reduce motion artefacts

- 21 cardiac phases
- 3C data acquired in 20-22 sagittal slices aligned with the aortic arch
- FOV = 280x280 mm
- Isotropic spatial sampling (Voxel Size 2x2x4 mm, slice spacing 2 mm)

Helical Blood Flow in the Human Aorta

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A better understanding of the role of pitch and torsion in blood flow development can be obtained through **helicity**, a scalar eligible to study relationships between complexity and energy.

Roughly speaking, **helicity gives measure of alignment of velocity and vorticity**

Like energy, **helicity influences evolution and stability of both turbulent and laminar flows** (Moffatt and Tsinober, 1992).

Helicity related to the **reduction** of non-linear processes responsible for transfer and redistribution of energy through various scales, and hence **energy dissipation**
Particle traces computed by time integration of the velocity field
(4th-order Runge-Kutta)

Sets of $N_p$ immaterial particles released at 5 different phase in systole
and tracked up to end systole. No real tracer used.

Bicubic spline interpolation both in the spatial and time domain.

FDM implemented for velocity gradients calculation

Accuracy algorithms tested on synthetic 4D flow data mimicking a virtual
PC-MRI acquisition (Morbiducci et al., 2009; Ponzini et al., 2009; Morbiducci et al., 2011).
In Vivo Quantitative Helical Blood Flow - the First Study

Morbiducci et al. *Annals of Biomedical Engineering* 2009

Helical Blood Flow in the Human Aorta Quantified by PC - MRI

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Helical Flow Index - HFI

begins with:

\[ H_v(s; t) = V \cdot (\nabla \times V) = V(s; t) \cdot \omega (s; t) \]

\[ \text{LNH}(s; t) = \frac{V(s; t) \cdot \omega (s; t)}{|V(s; t)| \cdot |\omega (s; t)|} \quad -1 \leq \text{LNH} \leq 1 \]

ends up with:

\[ \text{HFI} = \frac{1}{N_p} \sum_{k=1}^{N_p} \frac{1}{(t_k^{\text{end}} - t_k^{\text{start}})} \int_{t_k^{\text{start}}}^{t_k^{\text{end}}} |\text{LNH}_k(\zeta)| d\zeta = \frac{1}{N_p} \sum_{k=1}^{N_p} hfi_k \quad 0 \leq \text{HFI} \leq 1 \]

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Results – Acquired PC MRI Data

3C velocity map frames (phase I, II, and III) on a plane aligned with the aortic arch, viewed from the left. Brightness is proportional to signal intensity.

Helical Blood Flow in the Human Aorta

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Anatomical reconstruction of the aortas, together with the measured blood flow rate waveforms.
Evolution of particles sets emitted at early systole
Blood is conveyed into the aorta with streaming patterns aligned with the aortic axis: no formation of evident helical vortices can be appreciated
Evolution of the particle set emitted after peak systole is strongly characterized by the onset of more coherent helical structures.
4D Evolution of the Aortic Flow – SUBJECT C

Helical Blood Flow in the Human Aorta

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Helical Blood Flow in the Human Aorta

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The flow deceleration phase is dominated by the fluid rotational momentum, resulting in coherent helical and bihelical patterns displaying the first 25 ms of motion of particle sets. Image is oriented as if observer is looking inferiorly.
Helical Flow – Quantitative Analysis I

**INTRAINDIVIDUAL ANALYSIS**

**features common to all:**
- Particle sets emitted after peak-systole, highest helical content
- Particle sets emitted during acceleration phase characterized by similar trends in HFI values

**bulk flow helical content depends upon the evolution of the flow through the aorta**

*Helical Blood Flow in the Human Aorta*

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INTERINDIVIDUAL ANALYSIS

mean HFI values

very similar values of mean HFI

healthy individuals exhibit characteristic average systolic content of helical blood flow in aorta
Conclusion

There were two key findings of our study:

(i) intra-individual analysis revealed a statistically significant difference in the helical content at different phases of systole

(ii) group analysis suggested that aortic helical blood flow dynamics is an emerging behavior that is common to normal individuals

The ability to apply the technique to assess helical blood flow in vivo could be helpful to raise to still unanswered questions concerning the primary circulation.

Its ability in ranking fluid dynamical behaviour candidates HFI for diagnostic use in clinical practice.

Our results enforce the hypothesis that helicity contribute to optimize the naturally occurring fluid transport processes in the cardiovascular system, aiming at obtaining an efficient perfusion, avoiding excessive energy dissipation in the process of conveying blood flow in aorta.
Helical Blood Flow in the Human Aorta
On the Use of In Vivo Measured Flow Rates as Boundary Conditions for Image-Based Hemodynamic Models of the Human Aorta. Implications for Indicators of Abnormal Flow


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Flow induced wall shear stress (WSS) is thought to play an important role in the initiation and progression of vascular diseases. Accurate assessment of WSS in aorta is of paramount importance in order to get further insight into the comprehension of the role played by WSS in vascular disease.

However, while in vivo direct measurements of blood velocities in the bulk and flow rates in aorta are sufficiently affordable and accurate, reliable in vivo estimation of WSS is still a challenge.

Coupling medical imaging and CFD allows to calculate highly resolved blood flow patterns in anatomically realistic models of the thoracic aorta, thus obtaining the distributions of WSS at the luminal surface.

However, the increasing reliance on CFD for hemodynamic simulations requires a close look at the various assumptions required by the modeling activity.

In particular, much effort has been spent in the past to assess the sensitivity to assumptions regarding boundary conditions (BCs).
(1) to identify the individual, not invasively measured PC MRI-based BCs scheme that better replicates the measured flow rate waveforms;

(2) to describe the impact that different strategies of combining PC MRI-based outlet BCs have on WSS distribution. The identification of a proper set of individual not-invasively measured BCs can eliminate potential sources of error and uncertainties in blood flow simulations in the human aorta.
Rationale, Aims, How - Flow Chart

How

4D PC-MRI

Subject-specific models reconstruction

Mesh

Measured flow rate waveforms

Mesh sensitivity analysis

CFD simulations

Post-processing

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Methods

1) 4D PC-MRI \textit{in-vivo} data

Flow rate waveforms

\begin{align*}
    \frac{\partial u}{\partial t} + (u \cdot \nabla)u &= -\frac{1}{\rho} \nabla p + \nu \Delta u \\
    \nabla \cdot u &= 0
\end{align*}

a) - Navier-Stokes & Continuity

- blood:
  \begin{align*}
    \rho &= 1060 \text{ kg/m}^3 \\
    \mu &= 0.0035 \text{ Pa s}
  \end{align*}

b) UDF: Boundary Conditions

c) Export WSS on ASCII file

2) CFD

FINITE VOLUME (FLUENT SOLVER)

3) MATLAB

Post processing: TAWSS, OSI, RRT
Methods

Fluent Code settings:

- Velocity: second order upwind
- Pressure: linear interpolation
- Pressure –velocity coupling: SIMPLE
- Transient formulation: $\Delta t = 0.001$ ms
Subject-Specific Model Reconstruction

Model A1

- PC-MRI reconstructed human thoracic aorta;
- Hexahedral mesh of 1.5 million cells.

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Subject-Specific Model Reconstruction

Model A2

- PC-MRI reconstructed human thoracic aorta;
- Hexahedral mesh of 1.5 million cells.
- pyFormex: [http://www.pyformex.org](http://www.pyformex.org)
Measured Flow Rate Waveforms as Boundary Conditions in Hemodynamic Simulations

AAO – ascending aorta
Dao – descending aorta
BCA – brachiocephalic artery
LCCA – left common carotid artery
LSA – left subclavian artery

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## Boundary Conditions

- **P**: Stress free condition
- **COR**: Constant Outflow Ratio (% of AAo inlet flow rate)
- **MFR**: Measured Flow Rate

### Outlet Treatment Scheme

<table>
<thead>
<tr>
<th>Outlet Treatment Scheme</th>
<th>DAo</th>
<th>BCA</th>
<th>LCCA</th>
<th>LSA</th>
</tr>
</thead>
<tbody>
<tr>
<td>I</td>
<td>P</td>
<td>COR</td>
<td>COR</td>
<td>COR</td>
</tr>
<tr>
<td>II</td>
<td>MFR</td>
<td>P</td>
<td>P</td>
<td>P</td>
</tr>
<tr>
<td>III</td>
<td>P</td>
<td>P</td>
<td>P</td>
<td>P</td>
</tr>
<tr>
<td>IV</td>
<td>MFR</td>
<td>COR</td>
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</tr>
<tr>
<td>V</td>
<td>MFR</td>
<td>MFR</td>
<td>P</td>
<td>P</td>
</tr>
<tr>
<td>VI</td>
<td>P</td>
<td>MFR</td>
<td>MFR</td>
<td>MFR</td>
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</tbody>
</table>

### Constant Outflow Ratio

<table>
<thead>
<tr>
<th></th>
<th>Model A1</th>
<th>Model A2</th>
</tr>
</thead>
<tbody>
<tr>
<td>BCA</td>
<td>13.4%</td>
<td>26.7%</td>
</tr>
<tr>
<td>LCCA</td>
<td>10.6%</td>
<td>5.5%</td>
</tr>
<tr>
<td>LSA</td>
<td>12.0%</td>
<td>0.3%</td>
</tr>
</tbody>
</table>

Flow rate at AAo inlet section prescribed in terms of flat velocity profile.
WSS-based Descriptors of Abnormal Flow

**TAWSS** (Time Averaged WSS)

\[
TAWSS = \frac{1}{T} \int_{0}^{T} |WSS(s,t)| \, dt
\]

- TAWSS < 0.4 Pa \rightarrow atherogenic risk
- TAWSS > 1.5 Pa \rightarrow atheroprotective
- TAWSS > 10-15 Pa \rightarrow endothelial damage
WSS-based Descriptors of Abnormal Flow

OSI (Oscillating Shear Index)

\[
OSI = 0.5 \left[ 1 - \frac{\left| \int_0^T WSS(s, t) \, dt \right|}{\int_0^T |WSS(s, t)| \, dt} \right]
\]

- High OSI \quad \rightarrow \quad \text{intimal thickening}
**WSS-based Descriptors of Abnormal Flow**

**RRT** (Relative Residence Time)

\[
RRT = \frac{1}{(1 - 2 \text{OSI}) \text{TAWSS}} = \frac{T}{\int_{0}^{T} WSS(s,t) \, dt}
\]

- High RRT ➔ **atherosusceptible**
- Low RRT ➔ **atheroprotective**
Root Mean Square (RMS) of TAWSS, OSI and RRT was computed over patches.
Aorta models $A1$ and $A2$ were compared using Cohen distance $d$:

$$d = \frac{\mu_{A1} - \mu_{A2}}{\sigma_{pooled}} \quad \sigma_{pooled} = \frac{\sigma_{A1} + \sigma_{A2}}{2}$$

For a chosen index and a model $Ai$:

- $\mu_{Ai}$ is the area-averaged mean of the index;
- $\sigma_{Ai}$ is the area-averaged standard deviation of the index.
Mesh Sensitivity Analysis

Aorta A2: 6 refinements $10,000 \div 1,500,000$;

Distributions of descriptors associated with each grid were compared through Cohen $d$;

The trend of $d$ indicates that, refining the mesh, the descriptors become closer to the desired resolution.
Results – Computed vs Measured Flow Rates

DAO – *in-vivo* vs *in-silico* Flow Rate

scheme I (**blue**) - measured values approximated better for A2
scheme III (**red**) - highest differences
scheme VI (**light blue**) - agreement with *in-vivo* waveforms

![Graph showing flow rate comparison](image-url)
Results – WSS-based Hemodynamic Indicators

Neumann BC on DAo – P

TAWSS

(1) Proximal outer arch curvature
(2) Focal regions on DAo

TAWSS VI - Model A1
Results – WSS-based Hemodynamic Indicators

Neumann BC on DAo – P

TAWSS

(1) Proximal outer arch curvature
(2) Focal regions on DAo

TAWSS VI - Model A2
schemes II – III: imposition of stress-free condition at all the supra-aortic sections may reduce flow stagnation regions;

scheme I: on model A2, constant outflow ratio on LSA is 0.3% of the inlet flow at the AAo.
Results – WSS-based Hemodynamic Indicators

Neumann BC on DAo – P

OSI

Model A1

Model A2
Results – WSS-based Hemodynamic Indicators

OSI

Model A1

Model A2

scheme I: low OSI values on both models;
scheme III: on model A2, flow rate waveform of DAo has a damped dynamics with respect to other in-silico and in-vivo flow rate waveforms.
Results – WSS-based Hemodynamic Indicators

Neumann BC on DAo – P

RRT

Model A1

Model A2
**Results – WSS-based Hemodynamic Indicators**

**RRT**

**Model A1**

**Model A2**

**Scheme III**: high values on model A1, because of high OSI values;  
**Scheme I**: high values on model A2, as a consequence of low TAWSS values.
**Results – Interindividual Comparison**

**TAWSS** is always higher in model A1 (d > 0);

**RRT** is always higher in model A2 (d < 0);

**OSI** $d$ has positive or negative signs, depending on the BC scheme;

**Model A1 is more atheroresistant.**

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<th>III</th>
<th>IV</th>
<th>V</th>
<th>VI</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Cohen distance $d$</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>TAWSS</strong></td>
<td>0.3851</td>
<td>0.4246</td>
<td>0.6169</td>
<td>0.3652</td>
<td>0.3724</td>
<td>0.3666</td>
</tr>
<tr>
<td><strong>OSI</strong></td>
<td>0.0828</td>
<td>-0.1389</td>
<td>0.0004</td>
<td>-0.1179</td>
<td>-0.2032</td>
<td>-0.1601</td>
</tr>
<tr>
<td><strong>RRT</strong></td>
<td>-0.3196</td>
<td>-0.2131</td>
<td>-0.2252</td>
<td>-0.2395</td>
<td>-0.2734</td>
<td>-0.2655</td>
</tr>
</tbody>
</table>

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Conclusions

- Patient-specific hemodynamic simulations of aortic flow is feasible by applying scheme VI;

- Prescribing not-invasively measured flow rate as BCs on the supra-aortic branches and pressure on Dao (scheme VI):
  - in-silico blood flow rates match PC-MRI measurements;

- Different schemes of BCs can influence WSS-based descriptors:
  - they mainly affect descriptors value than their distribution;

It is recommended to prescribe time-varying outflow BCs based on in-vivo accurate measurements (for example VI).
On the Use of In Vivo 4D Velocity Profiles as Boundary Conditions for Image-Based Hemodynamic Models of the Human Aorta

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Rationale

Image-based hemodynamic models of cardiovascular districts can be sensitive to assumptions regarding boundary conditions.

Aim

Evaluate influence that velocity profiles prescribed at the inlet section (AAo) have in hemodynamic models of the human aorta on:

- Bulk flow
- Wall Shear Stress

How

Image-based hemodynamic models of human aorta & PC MRI individual measurements of 3D velocity profiles
Methods

Image-based model aorta

1 mesh 6 MLN tetrahedral cells

Finite Volume Method (Fluent Solver)
PRELIMINARY-STUDY

Steady state analysis
2 time frames (T1, T2)

Methods

Boundary Conditions
- 3D measured PC MRI Velocity profile
- Flat Velocity profile [V-mean measured (PCMRI)]
Methods – Inlet Boundary Conditions

Inlet section AAo
Velocity vectors

Flat V profile

3D PC MRI measured profile

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Results – Streamlines at T1

Flat V profile

3D PC MRI measured profile

Pathlines Colored by Velocity Magnitude (m/s)

May 03, 2011

ANSYS FLUENT 12.1 (3d, dpbns, lam)
Results – WSS at T1

Flat V profile

3D PC MRI measured profile

Contours of Wall Shear Stress (pascal)

May 03, 2011

ANSYS FLUENT 12.1 (3d, dp, pbns, lam)

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Methods – Inlet Boundary Conditions

Inlet section AAo
Velocity vectors

Flat V profile

3D PC MRI measured profile

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Results – Streamlines at T2

Flat V profile

3D PC MRI measured profile
Results – WSS at T2

Flat V profile

3D PC MRI measured profile

Contours of Wall Shear Stress (pascal)

May 03, 2011

UMANS FLUENT 12.1 (3d, dp, plns, lam)

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Conclusions

From preliminary analysis

- Inlet velocity profiles seem to influence both bulk flow and WSS distribution

WORK IN PROGRESS
Thank You for Your Kind Attention

Turin Biomechanics Ramblers

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