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Published in:
Biophysical Journal

DOI:
10.1016/j.bpj.2012.03.048

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Document Version
Publisher's PDF, also known as Version of record

Publication date:
2012

Link to publication in University of Groningen/UMCG research database

Citation for published version (APA):

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Download date: 05-03-2022
Molecular Structure of Membrane Tethers

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ABSTRACT Membrane tethers are nanotubes formed by a lipid bilayer. They play important functional roles in cell biology and provide an experimental window on lipid properties. Tethers have been studied extensively in experiments and described by theoretical models, but their molecular structure remains unknown due to their small diameters and dynamic nature. We used molecular dynamics simulations to obtain molecular-level insight into tether formation. Tethers were pulled from single-component lipid bilayers by application of an external force to a lipid patch along the bilayer normal or by lateral compression of a confined bilayer. Tether development under external force proceeded by viscoelastic protrusion followed by viscous lipid flow. Weak forces below a threshold value produced only a protrusion. Larger forces led to a crossover to tether elongation, which was linear at a constant force. Under lateral compression, tethers formed from undulations of unrestrained bilayer area. We characterized in detail the tether structure and its formation process, and obtained the material properties of the membrane. To our knowledge, these results provide the first molecular view of membrane tethers.

INTRODUCTION

Membrane tethers are long, thin tubes with walls constituted by lipid bilayers. They play important functional roles in eukaryotic cells and their organelles (such as the endoplasmic reticulum, mitochondria, and Golgi apparatus) (1–5). Membrane tethers are efficient structures for cellular transport and communication, as well as for storage of excess membrane area upon synthesis of new lipids or changes in cell volume (6–8). In vivo, tethers are pulled by molecular motors or induced by membrane deformation (9,10). Experimentally, tethers can be formed by application of a localized external force (e.g., using micropipette aspiration, optical or magnetic tweezers, or hydrodynamic drag) (11–16) or by compression of confined membranes (17,18). Using tether-pulling experiments in combination with theoretical models, investigators have been able to characterize a number of mechanical membrane properties (19–24) as well as lipid phase behavior (25,26). Theoretical models also can relate tether geometrical parameters to the pulling force and surface tension. However, due to the small diameters (below optical resolution) and dynamic nature of tethers, detailed structural information about tethers and their formation process is lacking.

Here we used molecular dynamics simulations to obtain molecular-level insight into the formation of tethers from model membranes. We employed the coarse-grained MARTINI model (27) to simulate lipid bilayer patches of 4608 and 18,432 dioleoylphosphatidylcholine (DOPC) molecules in water containing 500,000–3,700,000 water particles on a microsecond timescale. Large-scale simulations representing tens of millions of molecules warranted the use of a coarse-grained force field. MARTINI was previously applied successfully to simulate a wide range of lipid assemblies, including various lipid phases and their transformations, vesicle fusion, and monolayer folding (see Marrink et al. (28) for review).

METHODS

All simulations were performed with the Gromacs software package (v.4.5.4) (29). The MARTINI coarse-grained force field (27) was used. In this force field, molecules are represented by beads, each of which represents approximately four nonhydrogen atoms. DOPC is a standard component of this force field. The angle potential at particles C2 and D3 in the hydrocarbon chains was modified to an equilibrium angle of 100° and a force constant of 10 kJ/mol to better reproduce a decrease in the order parameter profile at the unsaturated bond.

We simulated two systems: a small (4608 lipids) and a large (18,432 lipids) DOPC bilayer solvated in ~500,000 and 1,400,000 water particles, respectively. In the large bilayer, the water amount was increased to ~3,700,000 particles at pulling forces of 100 and 200 kJ/mol nm to increase the box size in the direction normal to the bilayer plane.

For nonbonded interactions, we used the standard cutoffs for the MARTINI force field: the Lennard-Jones potential was shifted to zero between 0.9 and 1.2 nm, and the Coulomb potential was shifted to zero between 0 and 1.2 nm with a relative dielectric constant of 15. The time step was 20 fs with neighbor list updates every 10 steps. Lipids and water were coupled separately to a target temperature using the velocity rescaling thermostat (30) with a time constant of 1 ps. The system was coupled to normal and lateral pressures of 1 bar with a semi-isotropic pressure coupling scheme to provide a tensionless bilayer, or to a surface tension with a surface tension coupling scheme, in both cases using a Berendsen barostat (31) with a time constant of 4 ps and compressibility of 5 10⁻⁶ bar⁻¹. The simulation time was 2 μs for small systems and 1 μs for large systems. A summary of all simulations is given in Table 1.
TABLE 1 Summary of the simulations performed

<table>
<thead>
<tr>
<th>No. of lipids</th>
<th>Extraction conditions</th>
<th>Result</th>
<th>Tether parameters</th>
</tr>
</thead>
<tbody>
<tr>
<td>4608</td>
<td></td>
<td></td>
<td>r</td>
</tr>
<tr>
<td></td>
<td>( n_0 )</td>
<td>( f )</td>
<td>( \gamma )</td>
</tr>
<tr>
<td>4608</td>
<td>3</td>
<td>10</td>
<td>B</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>150</td>
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<td></td>
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<tr>
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<td>200</td>
<td></td>
<td></td>
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<tr>
<td></td>
<td>200</td>
<td>10</td>
<td>P</td>
</tr>
<tr>
<td></td>
<td>200</td>
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<td>T</td>
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<td>100</td>
<td>T</td>
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<td>100</td>
<td>T</td>
</tr>
<tr>
<td></td>
<td>300</td>
<td></td>
<td></td>
</tr>
<tr>
<td>18,432</td>
<td>3</td>
<td>1</td>
<td>B</td>
</tr>
<tr>
<td></td>
<td>10</td>
<td></td>
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<tr>
<td></td>
<td>100</td>
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<td></td>
<td>40</td>
<td>-4</td>
<td>T</td>
</tr>
<tr>
<td></td>
<td>-6</td>
<td>-6</td>
<td>T</td>
</tr>
</tbody>
</table>

Here B is bilayer bending, P is protrusion, T is tether, R is rupture, C is catenoid-shaped tether, F is bilayer folding (see text for details), \( n_0 \) is the radius of the lipid patch to which the pulling force is applied, \( f \) is the magnitude of the pulling force, and \( \gamma \) is the surface tension in the bilayer.

*Pulling force simulations: radius of the lipid patch to which external force is applied (with lipids restrained in lateral direction). Lateral compression simulations: radius of unrestrained patch (with restraints in the normal direction applied to the remaining lipids).

Further tether elongation was not possible due to the finite system size in the direction normal to the bilayer.

In simulations with an external pulling force, the center of mass of a lipid path of radius 3, 4, or 5 nm was pulled with a linear potential, resulting in a constant force. The lipids belonging to the patch were determined in the starting configuration. Position restraints in the lateral directions (\( x \) and \( y \)) were applied to the phosphate group of these lipids to keep them in the patch and prevent them from diffusing back into the bilayer. In the lateral compression simulations, position restraints in the normal direction were applied to phosphate groups of lipids at the bilayer perimeter, i.e., outside of a lipid patch of either 30 or 40 nm in radius, to prevent bilayer folding.

Tether diameters were calculated from the positions of the phosphate groups of lipids in the upper bilayer leaflet constituting the outer leaflet of the tether. We calculated the tether average radius (Table 1) considering only the part of the tether that adopts a cylindrical shape, i.e., disregarding the tip and the connection to the bilayer. The tether length was estimated from the maximum and minimum \( z \)-coordinates of the phosphate groups of lipids in the outer leaflet of the tether. Lipid flip-flop between the leaflets, given its low rate and the relatively short simulation times compared with typical flip-flop rates, was not taken into account in the calculation of the relative area strain between the leaflets and the nonlocal bending energy.

RESULTS AND DISCUSSION

Tethers were extracted from lipid bilayers by application of an external pulling force or by lateral compression. A summary of the simulations performed and their results is presented in Table 1. Tether formation was defined as bilayer deformation into a cylindrical shape (as opposed to bilayer bending and cone-shaped protrusions). Earlier simulations used preformed bilayer cylinders to calculate the bilayer bending modulus (32).

In the external force simulations, a constant force was exerted on the center of mass of a lipid patch in the direction normal to the bilayer plane. Coordinates of the phosphate groups of lipids in the patch were restrained in the lateral direction, modeling adhesion to a bead in experiments. Periodic boundary conditions in the lateral direction combined with the center-of-mass motion removal prevented normal displacement of the bilayer perimeter. A tether pulled from the large bilayer (80 x 80 nm\(^2\)) is shown in Fig. 1, \( a \) and \( b \). The tether geometry resembles a small nanotube: a narrow water channel is surrounded by a cylindrically curved bilayer of nearly constant diameter, widening at its connection to the membrane.

Tether development is illustrated in Fig. 1, \( c–e \), for the small patch (40 x 40 nm\(^2\)) under forces of different magnitude. A cone-shaped protrusion (Fig. 1 \( c \)) transformed into a cylindrical tube (Fig. 1 \( d \)), which grew in length (Fig. 1 \( e \)). Tether formation proceeded through these steps under strong pulling forces (cf. Table 1). Under weak pulling...
forces, only an initial protrusion formed. Even larger forces led to tube rupture, which was initiated by local thinning of its diameter (Fig. 2). The mechanism of rupture was similar to fission of liposomes (33).

The evolution of the tether radius profile as a function of time is shown in Fig. 3a. When the pulling force acted on larger lipid patches, the tube tip became wider than the tube diameter. This led to a theoretically predicted (34) catenoid-like shape (resembling the interior of a torus). This shape here is likely an artifact related to high pulling rates, because at lower pulling rates (in a larger bilayer) a cylindrical tube of a larger radius would be produced.

Theoretical models have described tether development as a two-step process including initial viscoelastic deformation of the membrane followed by viscous flow of lipids into the tube (35,36). The crossover between these two phases of tether development takes place for pulling forces above a threshold. In our simulations, we were able to reproduce this behavior. In Fig. 3b, tether development under a constant pulling force is shown. A crossover from a protrusion to a growing tether regime manifests as a change in slope in tether elongation versus time. The threshold force depended on the size of the bilayer, decreasing with increasing system size. Tether growth was linear at a constant applied force, in agreement with theoretical predictions (36). Tube linear elongation was always limited by the finite bilayer size in the simulation. Tether formation was also reversible: reduction or removal of the external force resulted in tether retraction or complete respreading. Application of a positive surface tension to the bilayer in combination with pulling force led to a reduction in the tether diameter, as expected, or, at higher magnitudes, prevented the transformation from protrusion to tether. Negative surface tension (i.e., bilayer compression), in turn, facilitated the flow of lipids into the tether.

Experimental studies (17,18) have shown that tethers can be formed by lateral compression of the membrane alone, i.e., without pulling. To mimic this in simulations, we applied a negative surface tension to large bilayers with position restraints at the perimeter to prevent bilayer folding (see Table 1). Bilayer folding was similar to monolayer folding (37) and resulted from bilayer bending in one direc-

![Figure 2](image1.png)  
**FIGURE 2** Tether rupture by a strong pulling force, shown in side (a), cross-section (b), and close-up (c) views. A force of 300 kJ/mol nm is acting on lipid patch of radius 4 nm in a DOPC bilayer of 4608 lipids. Color scheme as in Fig. 1; lipid tail ends are shown as pink spheres in panel c.

![Figure 3](image2.png)  
**FIGURE 3** Tether development under constant pulling force. (a) Distribution of the tether radius along its axis as a function of time. A force of 200 kJ/mol nm is acting on a lipid patch of radius 3 nm in a DOPC bilayer of 4608 lipids. The radius is determined using the coordinates of the phosphate groups in the outer leaflet. (b) Tether elongation as a function of time is shown for forces of 50 (circles), 100 (squares and triangles), and 200 (diamonds) kJ/mol nm acting on a patch of radius 3 nm in a DOPC bilayer of 4608 or 18,432 (triangles) lipids. The force of 50 kJ/mol nm is below the threshold, and the crossover from protrusion to tether does not occur (see text for details).

![Figure 4](image3.png)  
**FIGURE 4** Tether formation under lateral compression. Undulations of unrestrained bilayer area (a) are followed by the formation of a wider tether (b). A negative surface tension of −4 mN/m is acting on a bilayer of 18432 DOPC lipids. Position restraints are applied to prevent normal displacement of the lipids at the bilayer perimeter (unrestrained patch radius is 40 nm). Color scheme as in Fig. 1.

From the tether-pulling simulations we can extract material properties of the membrane, including the bending
modulus, viscosity, and tether tension. Static forces that
counteract tether elongation originate from local and non-
local bending energies associated with bilayer deformation.

The local bending energy arises from elastic resistance of
the bilayer to curvature as it is pulled into a tube. The
nonlocal bending energy accounts for the curvature-induced
relative strain between the leaflets of the closed bilayer
(containing a fixed number of lipids) (19,38). Using the
Helfrich Hamiltonian (39), the local bending energy reads
\[ E_b = \frac{\pi K_b}{r} \]
where \( K_b \) is the bilayer bending modulus, \( r \) is the
tether radius, and \( L \) is the tether length. Nonlocal bending
energy is given by (19,38)
\[ E_r = K_r \Delta A^2 / (2A_0 h^2) \]
where \( \Delta A = 2\pi L h \) is the relative area difference
between the inner and outer bilayer leaflets in the tube, \( K_r \) is
the nonlocal bending modulus, \( h \) is the distance between
the neutral surfaces of each leaflet, and \( A_0 \) is the total bilayer
area. It is worth noting that although the area of cells and
artificial liposomes (~1–100 \( \mu \)m\(^2\)) is large compared with
the area incorporated in the tether, its small value in our
simulations (~40 \( \times \) 40 or 80 \( \times \) 80 nm\(^2\)) for the two consid-
ered bilayer sizes) increases the contribution of this term
to the tether energy by several orders of magnitude. The
dependency of this term on the bilayer area leads to the
observed decrease of the threshold force with increasing
size of the simulated system (cf. Table 1). The nonlocal
bending term, which increases with tether length due to an
increase in the relative area difference between the leaflets,
also explains the limited linear tether growth at a constant
force in our simulations.

The tether force, which arises from both local and
nonlocal bending, is given by \( f = 2\pi K_b/r + 4\pi^2 K_r L/A_0 \).
Using a typical ratio (19,40) of \( K_b/K_r \sim 3 \), at a pulling force
of 100 kJ/mol nm and a patch radius of 3 nm, we obtain a
bending modulus \( K_b \sim 1 \times 10^{-15} \) J. The value is similar
for the small and large bilayers in a range of pulling forces
(cf. Table 2). A slight increase in the apparent \( K_b \) value with
increasing force likely originates from higher viscous
dissipation resulting in higher dynamic forces (at higher
pulling speeds). To calculate \( K_b \), we used the tether length
at the transition point between protrusion formation and
the beginning of the linear growth regime. The obtained
estimates are in good agreement with the experimentally
measured value of 0.9 \( \times \) 10^{-19} J (41) and previous simulation
result of 0.6 \( \times \) 10^{-19} J (42) for DOPC bilayers. Recent studies,
however, reported somewhat higher bending moduli for
bilayers in the MARTINI force field (43,44). Our value of
the bending modulus may be somewhat overestimated due
to the small radius of the tubes, leading to a possible
coupling between the bending and area stretching deforma-
tions not accounted for in the above formulas.

The surface tension in the tether, \( \gamma_r \), can then be obtained
with the following formula (19,22)
\[ r = \sqrt{K_b/2\gamma_r} \]
Using the above estimates, we find a surface tension in the
tether of ~1 mN/m, which is larger than the surface tension
in the tensionless bilayer, in agreement with theoretical predic-
tions (38). Note that the calculated bending modulus and the
surface tension in the tether increase with increasing pulling
rate. This is because the tether pulling forces used in the
estimates of these properties contain both static and viscous
friction contributions. Viscous dissipation in turn increases
with an increasing pulling rate.

The tether force that arises from viscous friction
comprises the bilayer surface flow, mutual slide of the leaf-
lets, and drag of the associated solvent (22,23,45). Viscous
friction can be assessed by calculating the effective surface
viscosity in the system. This effective surface viscosity re-
presents an apparent bulk viscosity multiplied by a charac-
teristic length, which is equal to the thickness of the
bilayer and the associated solvent layers. To quantify this,
we can assume a simple linear force-velocity relationship,
\[ f = f_0 + \eta_{eff} dL/dt \]
where the effective surface viscosity \( \eta_{eff} \) does not depend on the tether elongation rate \( dL/dt \). We
can then estimate the effective surface viscosity \( \eta_{eff} \) and the
force at zero velocity, \( f_0 \), by fitting this linear relationship to
the force versus growth rate data (cf. Table 2). Of interest,
the calculated static forces are close to the threshold forces
in our simulations. This explains why lower pulling forces
did not produce tethers in our simulations. The calculated
effective surface viscosities, ~1 \( \times \) 10^{-9} Pa \( \cdot \) m/s, are
in good agreement with experimental data for model bilayers
(22,23). Note that the calculated values decrease with
increasing size of the simulated system. This could result
from confinement of the associated solvent, which has
been shown to increase the apparent solvent viscosity
(46). Using the estimate for the bilayer surface viscosity
\( \eta_b \sim G(s)/s \) with the shear modulus \( G(s) \) at the pulling rate
s~0.1 nm/ns from Baukina et al. (37), we expect the
contribution of bilayer surface flow to the effective viscosity
to be relatively small at ~1 \( \times \) 10^{-10} Pa \( \cdot \) m/s. On the basis of
the combined results, we find that the contributions of static
and viscous forces to the total tether force are comparable.

In our simulations, several conditions are distinct from
pulling cellular tethers in experiments. Cellular membranes

### Table 2: Bilayer material properties calculated from tethers

<table>
<thead>
<tr>
<th>No. of lipids</th>
<th>( r_0 )</th>
<th>( f_0 )</th>
<th>( L_0 )</th>
<th>( K_b \times 10^{10} )</th>
<th>( \gamma_r )</th>
<th>( f_0 )</th>
<th>( \eta_{eff} \times 10^{10} )</th>
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<td>4608</td>
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<td>17</td>
<td>9</td>
<td>0.9</td>
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<td>5</td>
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<td>20</td>
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</tr>
</tbody>
</table>

Here \( r_0 \) is the radius of the patch to which the pulling force is applied; \( f \) is
the magnitude of the pulling force; \( L_0 \) is the tether length at the end of the
protrusion formation, which is followed by linear growth; \( K_b \) is the bending
modulus of the bilayer; \( \gamma_r \) is the surface tension in the tether; \( f_0 \) is the
tether force at zero velocity; and \( \eta_{eff} \) is the effective surface viscosity.
are typically attached to the underlying cytoskeleton; adhesion to the cytoskeleton contributes to the static force, and the flow of lipids through the cytoskeleton network leads to viscous dissipation (22,23,45,47). The crossover from a protrusion to tether elongation typically occurs when the membrane separates from the underlying cytoskeleton. In our model membranes, adhesion is not included in the pulling force simulations, and processes of a different nature contribute to both dynamic and static forces. Despite these differences, it is worthwhile to compare the tether extraction parameters in our simulations with experimental conditions. The forces that lead to tether growth in simulations (~100 pN) are one to two orders of magnitude larger than experimental values. At the same time, the tether radii lie at the lower limit (~10 nm) of the experimental values. Larger forces result in faster pulling rates, which in the simulations are at least two orders of magnitude higher than those used experimentally for model membranes (~10^{-4} m/s) (23). At comparable effective viscosities, faster rates lead to higher viscous forces. On the other hand, the static forces in the simulations are also larger than experimental ones due to the much smaller membrane size. The latter limits the scale of tether-induced deformation and thus leads to higher nonlocal bending energy, coupling between the bending and stretching deformations, and curvature dependence of the elastic moduli (48).

In summary, using molecular dynamics simulations, we reproduced tether formation from model membranes under applied external force as well as lateral compression. These simulations gave detailed structural information on tethers and tether formation, and provide a link between molecular simulation and membrane material properties.

This work was supported by the Natural Sciences and Engineering Research Council (Canada). D.P.T. is an Alberta Innovates Health Solutions Scientist. The simulations were performed on Westgrid/Compute Canada facilities.

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